Mediolateral Balance Assessment in older Adults: MELBA

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Mediolateral Balance Assessment in Older Adults: MELBA

ACADEMISCH PROEFSCHRIFT

ter verkrijging van de graad Doctor aan
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en de graad Doctor in de Biomedische wetenschappen aan de KU Leuven,
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door

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geboren te Temuco, Chile

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Dit proefschrift is tot stand gekomen op basis van een daartoe tussen de Vrije Universiteit en de KU Leuven, België, overeengekomen samenwerkingsverband ter regeling van een gezamenlijke promotie als bedoeld in het Promotiereglement Vrije Universiteit, hetgeen mede tot uiting wordt gebracht door de weergave van de beeldmerken van beide universiteiten op deze titelpagina.

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General Introduction

A matter of balance

It is expected that the percentage of older adults in the worldwide population will drastically increase in the next decades. In Western Europe and in countries such as Chile, China and Canada, the population over 60 years will represent over 30% of the total [1]. It is consensus that mobility limitations increase with aging, which combined with the growth of the older population will inevitably increase the burden on health care system. This urges the implementation of interventions aiming to prevent the occurrence of mobility disability with aging [2].

Mobility disability (MD) is defined as the inability to cope with daily-life mobility demands and has a profound impact on a person's quality of life, restricting social engagement and increasing dependency. Falls in the older adults may precipitate the occurrence of MD, especially when leading to serious consequences (i.e. hip fracture), which in some cases lead to prolonged bed rest and death. Some preclinical signs of mobility limitations are slower gait speed and reduced muscle power [3]. It is likely that impairments of multiple systems functioning contribute to falls and mobility limitations with aging of which balance plays a major role [4]. Balance impairment has been identified as a major contributor to the increased number of falls in the elderly [5] (Figure 1.1).

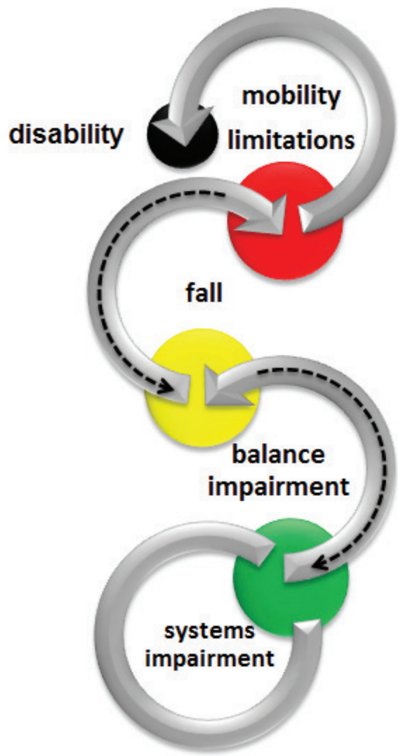

Figure 1.1. Diagram showing the path towards mobility disability initiated by multi-systemic impairment that leads to balance impairment. The green dot inserted represents the point in time for early detection of balance detriment, the yellow one the time at which inability to cope with daily-life perturbations may lead to a fall, the red one the time at which the latter has serious consequences for mobility and health and the black dot the time at which disability is diagnosed. Dashed arrows represent bi-directionality of cause-effect relationships between mobility limitations and falls and of the latter with balance impairment.

Balance control is highly challenged during locomotion (i.e. walking) and transitions (i.e. rising from a chair), activities during which an older person is more prone to fall [6]. Although aging is associated to decreased physical capabilities and increased risk of falling, behavioural and environmental factors are also important [7]. For instance, in relatively healthy older adults, falls may occur only when coping with perturbations in challenging environments whereas in frail peers this may occur when coping with small perturbations [8]. Since exposure to stability-threatening situations occurs on a regular basis, i.e. when traveling on public transportation or walking on uneven terrains, subtle impairments of balance may cause falls even in healthy community-dwelling older adults [8-10].

Although no consensus has been reached yet, impairment of mediolateral (ML) balance specifically has been associated to an increased risk of falling, based on posturographic and gait stability measures. In spite of not being directly associated, impairment of ML balance may partly explain falls due to incorrect weight-shifting as seen when qualitatively assessing the circumstances of a fall in health-care facilities [6]. The ability to shift weight is strongly deteriorated in older adults with neurological disorders such as Parkinson's disease [11] and stroke survivors [12], who have a high risk of falls.

Early detection of balance impairment, would be a first step to identify older adults at risk of falling who may benefit from preventive interventions aimed to maintain or improve balance control [13]. Unfortunately, few assessment tools are sensitive enough to determine subtle changes and even fewer specifically target ML balance. The aim of this thesis is to develop and test a new method for ML balance assessment, which may be more suitable than current tools for screening community-dwelling older adults with subtle impairments of balance.

The effect of sensory, motor and cognition detriments with aging on balance control.

Balance is the ability to maintain stable motion in spite of the pull of gravity, through sensorimotor integration constrained by task (e.g.. walking, upright standing) and contextual demands (e.g. uneven terrain, surface motion). Stability here refers to continuation of task performance without injury of any part of the body or segments involved [14]. Balance depends on the integration of multiple sensory systems, namely visual, vestibular, proprioceptive, somatosensory systems and communicating pathways. Also adequate functioning of the musculoskeletal system, which executes the motor commands according to the demands of the task and context is crucial to maintain stability. All mentioned systems, however, have been shown to be affected by aging [15-17]. In addition, dual-tasking inability has been also associated to lower balance capacities which indicates that cognitive resources may also play a role in maintaining stability [18]. The following sections will discuss on the effect of aging on the mentioned systems and their consequences for balance control.

Age-related changes in sensory systems contributing to balance control.

Vestibular system

The vestibular system has a craniocentric reference frame [19,20] that allows detecting deviations from the vertical [21]. Linear and angular accelerations are sensed by the macula receptors (utricule and saccule) and the receptors at the crista ampullae (semicircular canals), respectively [22]. Afferents of the vestibular nerve reach the vestibular nuclei, occulomor nuclei, cerebellum (flocculonodular node) and brain cortex (parieto-insular vestibular and sensorimotor cortices)[23] serving for vestibulo-ocular and vestibulospinal reflexes and also for sensorimotor integration [24,25].

Impairments of the vestibular system, studied through vestibulo-ocular reflexes in the older adults, have frequently been reported [26-28]. Although, no clear association between age-related vestibular dysfunction and balance deterioration has been established yet [16], vestibular dysfunction is associated with an increased risk of falling [29]. A recent study also shows an age-related decrease in vestibulospinal reflexes gain coupled to overcompensation of dorsiflexors muscles when applying galvanic vestibular stimulation. These responses were attributed to a higher level processing [30]. Furthermore, it is possible that vestibulospinal reflexes involved in motor responses during sudden perturbations are affected by ageing. Welgampola and Colenbatch (2001) suggested that vestibulospinal reflexes may be "part of a backup postural adjustment" that is facilitated when large or sudden perturbations are not properly sensed by other systems [31].

Age-related impairment of vestibular afferents is similar to those exhibited by bilateral vestibular hypofunction patients [32] for which rehabilitation therapy aimed to improve remaining vestibular input, has shown to be effective [33]. However, in cases of vestibular loss, rehabilitation aims for substitution by enhancing the use of other sensory sources [33]. Under absence of vestibular inputs, reliance on vision to maintain postural stability increases in both static and dynamic conditions and this is coupled with an increased energetic cost [34], possibly due to increased muscle activation. The latter mechanism may also serve to enhance spindle afferent inputs (proprioception) [35].

Visual system

The afferents of the optic nerve reach the striate cortex (occipital area) where visual information is interpreted (Prasad, 2011). Information related to the environment (i.e. proximity to objects) is used by the balance control to determine a global reference system [19]. The visual and vestibular systems are closely related through the vestibuloocular reflex, which stabilizes images on the retina during head rotations [36].

The contribution of vision to balance control has been explored through comparing posturography between eyes open and eyes closed conditions. Overall, these studies have found increased postural sway when closing the eyes, indicating that the contribution of visual information is important for balance control [37]. Furthermore, manipulation of this system, using oscillating environments, induces in-phase sway [38], which is more pronounced in older adults [39].

Low contrast visual acuity, contrast sensitivity [40] and depth perception [41] as well as specific visual pathologies [42] have been reported to be associated to falls and lower mobility performance [43] in older adults living in the community. This visual impairment reduces awareness of environmental threats increasing exposure to perturbations beyond balance capabilities [44]. Furthermore, when compared to sighted older adults, low-vision and blind older adults exhibit strongly impaired balance control [45].

Although vision impairment affects balance, visual training can partly reverse this effect. The inclusion of collision avoidance games into more traditional strengthening programs improved older adult's performance in an obstacle course, which was interpreted as improved balance due to vision training [46]. Another study found that Tai-Chi can improve balance in older adults with poor vision by enhancing residual capacities or up-weighting sensory information from vestibular and proprioceptive inputs [47].

Proprioceptive system

The proprioceptive system is distributed over the body and comprises muscle spindles, tendon organs, cutaneous and joint receptors [48]. Information arising from these receptors has been proposed to be processed in subcortical structures such as the right putamen [49]. However, the two components of proprioceptive information, joint position and movement sense, seem to be processed separately [48].

Aging strongly affects the morphological and neurophysiological properties of the receptors contributing to proprioceptive inputs [15]. Since the main source of proprioceptive information are the muscle spindles [50], muscle-tendon properties changes due to aging, such as loss of sarcomeres in-series and of tendon stiffness [17,51], may affect their responses to changes in muscle length [50]. In addition, direct denervation and structural changes of muscle spindles have also been proposed as mechanisms underlying impaired proprioception in older adults [50]. Age-related degradation of primary endings in muscle spindles, which code velocity of length changes, has also been shown in animal models [52]. These processes may occur in parallel to the loss of muscle strength explaining their strong association in aging [53].

Proprioception measured at the hip using joint position sense (JPS) and kinaesthetic tests showed that this system is significantly affected by ageing [54]. Proprioceptive inputs arise from many muscles involved in movement simultaneously which may explain why JPS at the ankle joint alone did not discriminate between faller and non-fallers [55]. However, in diabetic subjects with peripheral neuropathy who exhibit more severe proprioceptive impairments, kinaesthetic tests showed ability to predict falls when used together with strength measures [56].

Somatosensory system.

A wide variety of skin receptors characterize the somatosensory system which is able to provide proprioceptive (i.e. Ruffini's endings detecting skin stretch) as well as exteroceptive information (i.e. Merkel disc detecting skin pressure) [22]. Most of the somatosensory receptors are affected by aging either by reductions in the number or structural changes [15]. For instance, tactile sensitivity [57] as well as vibrational perception thresholds at the foot [57,58] are higher in older adults compared to young. Furthermore, lower scores in the two-point discrimination test at the toe were found in fallers than in non-fallers [59]. These age-related changes have been associated to an increased torque variance around the ankle and reduced postural control adaptation when vibration on calf muscles is applied [57]. It is likely that reduced background discharge activities of these receptors (mainly type I fibers) under weight-bearing conditions may be responsible for these changes in the elderly [60,61]. Further evidence shows that in unconstrained standing, older adults produced less weight-shifting than young peers which has been suggested to be associated to detriment of the somatosensory system with ageing [62].

When standing, cooling of the feet soles increases medium latency reflex responses of calf muscles elicited by applying vestibular manipulations, highlighting the importance of feet sole receptors in modulating reflexes used for balance control [63]. Another study using anaesthesia on the foot soles found motor strategies changes when subjects were subjected to ML ground perturbations exhibiting greater reliance upon hip musculature to maintain stability [64]. These studies highlight the contribution of the somatosensory system to balance control. However, the use of vibration receptors (Meissner's and pacinian corpuscles) for balance, as largely affected by aging, is still to be explored [15].

Age-related changes in the muscular system associated to balance impairment

From middle age on, the musculoskeletal system loses muscle mass, which occurs at an increasing rate with ageing, eventually leading to sarcopenia, with a predominant loss of type II fibres [65,66]. Muscle mass can be reduced by 40% around the 8^{th} decade of life [67], while strength and power are even more affected Muscle architecture has been also shown to be affected by aging with shorter fascicle lengths (reduced sarcomeres in-series) and smaller pennation angles [68]. These changes coupled with reductions in parallel sarcomeres and increases in tendon compliance are likely to affect mechanical properties of the muscle (length-tension and force velocity relationships) [17].

Age-related impairment of the musculoskeletal system strongly affects balance performance. Reduced quadriceps [69], ankle plantarflexors [70] and dorsiflexors [71], hip abductors [72] and whole leg strength [73] have been associated to an increased risk of falling. Furthermore, reduced muscle power has been associated to increased postural instability [74] and lower performance on functional tasks such as rising from a chair and walking [75,76]. Besides, muscle weakness in older subject has been associated to lower ankle moments used to maintain stability when tripping [77]. Fortunately, resistance training has shown to partially reverse balance impairments [78,79] and may improve abilities to deal with daily-life sudden perturbations [80].

Age-related changes in sensorimotor integration

When coping with daily-life perturbations such as surface disturbances when commuting, being pushed on a busy street or tripping over an obstacle, sensory information from all available sources is rapidly processed at peripheral and central levels. This process is called sensory integration. In this process, sensory inputs are prioritized according to availability and reliability, which is called multisensory (re)weighting [81,82]. Balance, at a higher processing level, involves several structures of the brain and cerebellum which process and filter sensory inputs before executing a motor command [25]. These motor outputs finally rely on neuromuscular functioning for correct execution of motor responses. The whole process that comprises sensory integration, sensory reweighting and motor responses within a given task is known as sensorimotor integration [83].

The main source of inputs utilized to maintain stability when standing quietly comes from proprioception and vision [84] with small contributions from feet cutaneous receptors [85] and the vestibular system [86]. Studies using manipulations of visual, proprioceptive (muscle spindles and cutaneous receptors) [87] and vestibular [88] systems have shown that the balance control system is able to down-weight erroneous inputs and up-weight reliable ones to maintain stability. This sensory weighting process, however, has been shown to be affected by ageing, increasing the weight of visual information when standing still [89,90] possibly due to deterioration of proprioception. Possibly this also explains why older adults are more sensitive to visual disturbances [39].

The proprioceptive system is the most important to control upright stance (even when up-weighting visual inputs)[35,91] yet it is the sensory modality most affected by ageing [50]. Hence manipulations of this system using compliant surfaces [91], muscle tendon vibration [90,92], sudden ground translations [93,94] and sway-referencing pitch rotations [95,96] have been shown to have a larger effect on balance control in older than in young adults. In contrast, a study using visual and touch (fingertip) sensory manipulations during tandem stance found that the reweighting process was not affected by ageing, not even in fall-prone subjects [81]. This may reflect a distal-proximal proprioceptive deterioration that may maintain use of upper limb sensory information [15]. The distal-proximal hypothesis refers to the degradation initiated at the distal regions of the lower limbs (feet) that progresses towards proximal (hips) [15].

Older adults exhibit increased utilization of more proximal musculature in order to maintain stability. For instance, a greater demand on hip musculature has been observed during upright stance, which is accentuated in narrower base of support conditions [97] and when applying ground perturbations [98,99]. Distal-proximal reductions in sensory information [15] coupled with the same pattern of motor neuron loss [100] may partly explain such age-related sensorimotor changes.

Most studies analyzing the effect of ageing on balance control use postural sway measures during sensory manipulations as an indicator of balance control. However, it is possible that increased postural sway is the reflection of compensatory mechanisms aimed to increase somatosensory information (exploratory behaviour) [101]. Age-related proprioceptive and somatosensory deterioration at the foot may elicit this exploratory behaviour to increase the gain of these inputs for balance control. In fact, a previous study has shown that due to similar neural encoding, cutaneous and muscle afferents may compensate each other's errors during ankle joint movements [102]. This mechanisms may also explain postural sway increases in blind subjects to enhance sensory inputs for postural balance [103]. Sensory weighting has been also explored in more dynamic conditions such as walking. By using manipulations of the vestibular system with galvanic stimulation, Deshpadne and Patla showed that older are less able than young adults to down-weight erroneous inputs, as reflected in larger path deviations [104]. Another study also showed that older adults are less effective in down-weighting visual inputs when walking under visual perturbation conditions than young adults [105].

Sensory and motor cortices appear to be more affected than other cortical areas by ageing, which is reflected by grey matter thickness reduction [106]. Furthermore, it has been shown that integrity of white matter is also affected by aging, which was associated to lower sensory weighting abilities and lower balance performance [107]. Age-related changes in the structure of the right putamen have been associated to impaired proprioception [49] which, coupled with reductions in the number of myelinated fibres [108] and conduction velocity in peripheral nerves [109], can affect the dynamics of sensory weighting. This is also likely to affect balance responses when dealing with sudden changes of environmental stimuli due to slower conduction (peripheral nervous system) as well as processing (central nervous system) of sensory inputs [110,111]. Reorganization of the motor cortex [112] as well as a reduction in the number and increase in the size of motor units due to re-innervation [113], may also impair the sensorimotor integration process [114]. These changes may increase co-activation of agonist/antagonist muscles, which is thought to subserve increasing stability in the elderly [115-117], but which may, on the other hand also, decrease stability by reducing flexibility to respond to unexpected perturbations [118].

Age-related changes in cognition and balance control.

It has been proposed that not only physiological but also cognitive resources are utilized to control balance [119]. There is a larger demand on cognitive resources to maintain stability in older adults [120], especially when increasing the balance task complexity (i.e. simultaneous dual-task and sensory manipulations) [121]. Furthermore, performance on dual-tasks, such as walking and performing reactive or visuospatial tasks was shown to be associated to history of recurrent falls [122]. Although affected by concurrent cognitive tasks, older adults prioritize balance over performance on the double task, which otherwise may threaten stability and increase the risk of falling [123]. However, it is still not well known to what extent inability to perform in dual-tasks can predict falls [124]. Although both cognitive and motor performance in dual-tasks can be improved by dual-task training in the elderly, it is not known whether results can be translated into daily-life balance performance [125].

Assessment of balance in older adults

Balance assessment has frequently been proposed as a screening tool for increased risk of falling and mobility impairments in older adults population [5]. This because deterioration of balance control is the reflection of failure of multiple systems to different extents, which ultimately affects the abilities to cope with daily-life physical challenges [126]. Although it has been suggested that the association between balance impairment and risk of falling is moderate [5], it must be noted that several clinical tools show ceiling effects when used in community-dwelling older adults [127,128] likely underestimating the strength of this association. In fact, clinical tools should be used in accordance with the population to be assessed and settings in which they will be applied [129]. Balance impairment may not be clinically evident in early stages, which may hamper early diagnosis and potentially expose improperly diagnosed individuals to an increased risk of falling [130]. For instance, the timed-up-and-go has been suggested as a strong predictor of falls, however, it does not discriminate between fallers and non-fallers among highly functioning older adults [131].

Laboratory and computerized measures of balance have also been proposed to determine impairments and increased risk of falling in relatively healthy older adults. Most of these tests use static and dynamic posturographic measures (centre of mass and/or centre of pressure) [132]. More challenging tests including mechanical perturbations and eliciting stepping responses, as well as gait stability measures have also been suggested as tools for balance assessment. The following subsections will discuss the recent evidence on the use of clinical or laboratory and computerized systems to assess balance in community dwelling older adults.

Clinical assessment of balance

Clinical balance assessment tools can be divided in those that functionally assess balance and those that have a more physiological approach to identify the cause of balance problems [133]. The Berg balance scale (BBS) [134] and performance-oriented mobility assessment-balance section (POMA-B) [135] are probably the most used in geriatric settings. However, these tests exhibit ceiling effects when used in community–dwelling older adults [128]. Besides, most clinical tests do not consider the reactive component of balance control [136]. The latter is highly important when dealing with unexpected challenges for balance control occurring in daily-life situations [127] and it has been proposed that this should be assessed separately [137].

Clinical tools such as the physiological profile assessment (PPA) [138] and the balance evaluation systems test (BESTest) [139] aim to identify balance sub-systems impairments [133]. For instance, the PPA assesses vision (acuity and contrast sensitivity), vestibular function (visual field dependency), proprioceptive function (tactile, vibration and JPS) and muscle strength (knee flexors/extensors and ankle dorsiflexors), separately, as well as reaction time (hand and foot reaction to light stimulus) and postural sway (on a firm and compliant surfaces) [138]. Although the PPA is a comprehensive test, the validity of the test has been questioned mainly because postural sway measures misclassified fall risk in 58% of subjects [140]. The BESTest uses a combination of previously validated tools i.e timed up-and-go [141] and functional reach [142] with functional assessments of different sensory systems contributing to balance control (i.e. anticipatory and reactive postural responses) [139]. A shorter version of the latter test has also been developed (mini-BESTest) from which all biomechanical (i.e. strength measures) and stability limit measures (i.e. reaching) are removed [143]. A recent study determined the clinical utility of the BESTest and mini-BESTest in healthy older adults categorized by age by determining normative scores [144], however, clinimetric properties of this test (i.e. ceiling effect) in community-dwelling older adults have not been studied.

The construct of most clinical tests is based on single or multiple items that replicate situations occurring in daily-life such as rising from a chair or walking. For a detailed review on this issue please refer to [127]. The vast majority of these tests have been conceived to be used in clinical populations to diagnose balance problems, quantify balance impairment, guide treatment and determine prognosis [13]. However, their use has been extended into community-dwelling older adults without apparent problems, leading to low sensitivity and responsiveness of the tools [128]. In this regard, it has been suggested that early detection of balance impairments in able-bodied older adults requires new methods of a greater complexity, so as to replicate the underlying challenges of daily-life situations [127].

Laboratory and computerized measures of balance

Static Posturography

Forceplates and inertial sensors are used to obtain posturography measures, CoM and/or CoP sway, from which several parameters are calculated to quantify, for example, the amount and velocity of sway in mediolateral and anterior-posterior directions [145]. Although these methods provide insights in balance control at a group level, they have been criticized for their low utility for clinical decision making at the individual level [132]. This is mainly due to the fact that these methods are not able to differentiate among causes of balance problems [133,146]. However, they offer the advantages of easy application, relatively low-cost and low time-demands [145].

Overall, static posturographic studies have concluded that older adults exhibit impaired balance reflected in significant increases of sway when compared to young [147]. In a meta-analysis of posturography [148], measures of ML spontaneous sway amplitude with eyes open [9,149] , eyes open in a dual task [150] and eyes closed [151], root-mean-square (area ellipse) of the ML CoP excursion [152], minimum CoP velocity with eyes open [9] and mean sway velocity eyes closed [153] predicted or were associated with falls in older adults. Other studies, nevertheless, have found no evidence of such associations [154-156] perhaps due to the fact that upright standing may not reflect challenges as encountered in outdoor (daily-life) situations [157]. It has also been suggested that increased CoP-sway in the elderly reflects an effort of the balance system to compensate for sensory impairments (proprioception, visual or vestibular) by increasing somatosensory feedback [101,158]. Thus interpretations of aging effects on static posturography are ambiguous.

Dynamic posturography

Dynamic posturography can be subdivided into methods in which subjects are instructed to perform a movement task when standing, those in which sensory inputs are manipulated and those including mechanical perturbations. In the first group, perhaps the most clinically utilized is the limits of stability test (LOS), which determines the maximum excursion of the CoP when leaning in 4 or more directions. The LOS has been shown to be reliable when used in healthy older adults [159] and is also related to history of falls [160]. Nonetheless, similar reliability is offered by less sophisticated methods such as the multidirectional reach test [161]. Furthermore, LOS sensitivity to aging and prospective prediction of falls has not been determined.

Sensory manipulations can be used to determine whether a subject exhibits an increased reliance on one or more of the systems contributing to balance control, but also whether the balance control system is able to re-weight reliable and erroneous inputs [21,132]. Studies using specific manipulations for each sensory system are described above (section 1). The sensory organization test (SOT) is perhaps the most clinically used computerized methods using combinations of sensory manipulations (visual and proprioceptive), to assess reliance on and interactions of sensory inputs in older adults [132]. Although the SOT includes sway-referenced rolling, this is intended as a proprioceptive manipulation and not as a stability-threatening condition. Balance performance in this test has been shown to decrease with aging [162] predicting indoor yet not outdoor falls [153]. This test only considers aspects of balance in the sagittal plane in a standing position [127]. In summary, studies have found that older adults exhibit balance impairments associated with less adequate sensory reweighting, increased visual reliance and deterioration of proprioceptive and vestibular contributions to balance.

The use of mechanical perturbations allows determining whether a subject is able to utilize all the available resources in a stability-threatening situation and has been proposed as an alternative for clinical assessment of balance [99,163]. Anterior-posterior disturbances applied at a ground level have highlighted the importance of reactive and anticipatory mechanisms to control balance [164], revealing that although the first are affected by aging, this is not the case for anticipatory mechanisms [99]. These findings can be explained by an increased cortically-mediated control of balance in older adults [121,165]. The use of ground perturbations has been recently incorporated in more comprehensive balance assessment methods [166]. Although considered an advantage [132], standardized perturbation magnitudes do not allow determining the maximal capacities of a subject. A recent method, the closed loop system identification technique (CLSIT), uses sensory manipulations in combination with ground perturbations [166]. Further research is needed to determine its utility as a screening and/or fall prediction tool in healthy older adults.

Stepping Responses

Stepping responses are part of the reactive/protective repertoire to cope with perturbations that go beyond the capabilities to maintain the CoM within the base of support either when standing or walking [167,168]. Outcome of these tests is whether and how many steps are taken after a standardized perturbation. A prospective study showed that the use of multiple steps to maintain stability after a ML direction perturbation can predict falls in community-dwelling older adults [169]. Similar results were obtained by others when using multi-directional perturbations at CoM level, highlighting the importance of ML stepping responses for balance control [170]. The ability to maintain stability with a single step after a perturbation is likely affected by reduced hip joint torque generating capacities in the elderly [169,171]. Although important for mediolateral stability, the occurrence of stepping responses are a compensatory mechanism for when more efficient balance responses are not available (weight-shifting). This is supported by the fact that stepping responses are more likely to occur in those older adults that have experienced a larger number of falls [172].

Balance during gait

Since most people with and without evident balance disorders fall mainly when walking [173], balance assessment during gait is important in determining fall risk [174]. There is a vast literature reporting changes in gait with ageing, which are likely to reflect, in part, compensatory mechanisms of the balance control system to deal with the functional impact of multisystem impairments [175]. Besides spatiotemporal measures of gait (i.e. step length and width and walking speed) [176], laboratory measures can assess kinematic and/or kinetic parameters that provide more insight into the neuromuscular mechanisms underlying changes in gait with aging [177]. However, most of these measures are laborious to perform and demand high-cost technologies and technical expertise. On the other hand, the use of inertial sensors has emerged as an alternative for assessment of gait stability during treadmill and daily-life walking [178] and have been shown to be able to predict fall risk in older adults [179,180].

On the impairment of mediolateral balance in older adults

Several studies have concluded that age-related impairment of balance, mainly in the ML direction, has an important role in the ability to cope with daily-life physical challenges that may lead to falls [6,59,148,149,151,156,169,172,174,181-183] and hip fractures [151,184]. For instance, it has been concluded that the most common cause of falling is incorrect weight-shifting under different circumstances such as walking, standing or sitting down [6]. In line with this, prospective studies have found that the spontaneous CoP sway and gluteus medius activation onset after ML perturbations are strong predictors of falls in community dwelling older adults [151,156]. Furthermore in a review of posturography studies, although not conclusive, it was found that measures of ML balance have the most predictive value in older adults [148].

Balance during walking has been also found to decline with age when walking [185], with a narrow-base [186] and when crossing obstacles [185,187]. Furthermore, frequency analysis has shown greater power of ML movements at lower frequencies in older adults, suggesting slower weight-shifting than in young adults [188]. Taken together, the evidence indicates that impaired mediolateral balance control when standing and walking reflects weight-shifting impairments in common daily-life tasks, which may lead to an increased risk of falling.

As well as for overall balance, specific impairment in the mediolateral direction can be explained by impairment of sensorimotor mechanisms due to degenerative changes in the nervous and musculoskeletal systems. It is also possible that an inability to integrate left and right side sensory information has impact on weight-shifting abilities in the elderly. Asymmetric weight-bearing is a common manifestation in a variety of neurological conditions such as stroke [189] and Parkinson's disease (PD)[190]. Although to a different extent, PD [191] and aging [107] are associated to detrimental structural changes in the transcallosal motor tract, which is essential for interhemispheric interaction of the two motor cortices [191]. Therefore, It is possible that these changes may be a common path for the impairment of weight-shifting abilities suggesting this as an early indicator of balance impairment [190].

Despite of the importance of mediolateral balance, few tools specifically assess its impairment. Among those tests that do, three are commonly used: the one-leg-standing test (OLST), tandem stance (TS) and lateral functional reach (LFR) [156,192-194]. Scores in the OLST have shown to decrease with age [195], to be associated with decreased strength of the hip ab/adductors [196], hip flexor/extensors [197] and overall leg strength in a leg press machine [198]. Furthermore, OLST scores below 30 seconds have been associated retrospectively to an increased risk of falling [193]. The TS is considered a "static" measure of lateral stability for which performance decreases with age, which is related to decreased hip extensor and abductor strength [197] and is associated to an increased risk of falling [199]. A similar test, the near tandem stance (NTS) has been shown to be associated retrospectively to an increased number of falls, especially in those older adults unable to complete the test with eyes closed, as evidenced by taking sideward steps. Performance on this test was associated to decreased proprioception and muscle strength [172]. It is noteworthy that the TS test is mainly used as part of performance batteries such as the Berg balance scale and the short physical performance battery [200,201]. Finally, the LFR has been proposed as a reliable test of mediolateral stability in older adults [194] and has been related to the ability to perform daily-life activities [202]. However, no studies of sensitivity or predictive value for fall risk have been conducted.

Early detection of balance impairment, especially in the mediolateral direction is paramount to detect older adults who can benefit from preventive interventions before this evidently increases the risk of falling [145]. However, as previously discussed, most clinical balance assessment tools including those for the mediolateral direction, have exhibited low reliability, ceiling effects and low responsiveness when used in able-bodied older adults [127,128,203]. Therefore, several authors have made suggestions for the development of tools which in general terms should consider: assessment of reactive balance [204], greater complexity for ecological validity [127,132], more challenging when used in able-bodied older adults [128], use of computerized systems applicable to clinical settings [133], greater cognitive involvement [124], weight-shifting abilities [6], mechanical perturbations [132,145,163], inexpensive and easy application[132,145,163], and short duration [132].

Aim and outline of this thesis

The aim of this thesis was to develop and test a method that can quantify mediolateral balance impairment and that is suitable for highly functioning older adults who may exhibit subtle impairments of balance.

First, the feasibility of using visual feedback by determining whether enhancing visual information may mask impairment of other sensory sources contributing to balance control, was explored. In the study presented in Chapter II, it was investigated whether balance control can overcome erroneous sensory information arising from vestibular manipulation using transmastoidal galvanic current, proprioceptive manipulation at the hip muscles using muscle-tendon vibration and somatosensory manipulation using transcutaneous nerve stimulation at the feet soles when explicit feedback of actual body movement is provided.

A tracking task using the centre-of-pressure (CoP) as feedback on performance is presented in Chapter III: the mediolateral balance assessment tool (MELBA). In this chapter some methodological properties of the assessment tool were studied in a group of young adults. Since some concerns related to the overestimation of tracking capabilities as well as to the imposed CoM ML displacement were raised, the study presented in Chapter IV compared CoP and CoM feedback on performance during MELBA tasks. From this study, it is concluded that CoM tracking is more intuitive and more suitable to elicit consistent centre-of-mass (CoM) ML displacements, while allowing for a wider range of movement strategies. Therefore, an improved version of the method using CoM instead of CoP is presented in Chapter V. This study investigated the reliability of the test as well as sensitivity to age in comparison with more conventional measures of balance such as clinical tests (i.e. Berg balance scale) and conventional posturography. In the study described in Chapter VI, the ecological validity of the test was studied by determining its correlation with measures of gait stability, known to be predictive for falls, during treadmill and daily-life walking. Finally, in Chapter VII, general conclusions about the development of the test are presented. Suggestions for implementation and clinical utilization as well as further research using the tests are also discussed.

II

Can visual feedback efface the effects of sensory manipulations on mediolateral balance performance?

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Abstract

Visual feedback on performance is often used in balance assessment and training. However, due to the system's flexibility of reweighting sensory information for balance control, up-weighting of visual information may mask the detriment of other sensory systems. We therefore aimed to determine whether the effects of somatosensory, vestibular and proprioceptive manipulations on mediolateral balance are reduced by the presence of visual information and by explicit visual feedback on mediolateral sway of the body centre of mass. We manipulated sensory inputs of the somatosensory system by transcutaneous electric nerve stimulation on the feet soles (TENS), of the vestibular system by galvanic vestibular stimulation (GVS) and of the proprioceptive system by muscle-tendon vibration (VMS) of hip abductors. The effects of these manipulations on mediolateral sway were compared to a control condition without manipulation under three visual conditions: explicit feedback of sway of the body centre of mass (FB), eyes open (EO) and eyes closed (EC). Mediolateral sway was quantified as the sum of energies resulting from the power spectrum, at manipulations main content and at those not contained. Repeated measures ANOVA were used to explore the effect of each of the sensory manipulations, the effect of visual inputs conditions and their interaction on sway energy. Overall, all sensory manipulations increased body sway when compared to the control conditions for each of the visual input conditions with increases ranging from 6.8% at TENS-EO to 134% at VMS-FB. Absence of normal visual information had no effect on sway, while explicit feedback on body CoM reduced sway, although it did not efface the effect of manipulations. Furthermore, no interactions of visual information and sensory manipulation were found. In conclusion, this investigation highlights the potential of using visual feedback to assess and train balance since the use of feedback (up-weighting of visual inputs) on balance assessment cannot overcome and therefore mask sensory deficits.

Introduction

Balance in an upright posture is controlled by feedback based on visual, vestibular, proprioceptive (mainly from muscle spindles) and exteroceptive (in particular from cutaneous receptors in the soles of the feet) information [21,60,61,87,205]. However, the availability and reliability of these sensory inputs can be affected by intrinsic and environmental changes, which requires flexibility and adaptation of the balance control system [206-209]. This flexibility is achieved by reweighting sensory inputs based on reliability and pertinence to balance control in a given situation. For example, proprioceptive input from muscle spindles in the calf muscles contains limited information on orientation relative to vertical when standing on a compliant foam surface. In this situation, effects of manipulations of spindle inputs have been shown to be reduced compared to standing on a rigid surface. Similarly, vestibulospinal reflex gains are higher when visual inputs and tactile information are not available to the balance control system [31].

Sensory reweighting has been studied mainly by decreasing pertinence or reliability of information in one of the sensory channels involved in balance control. The effect of enhanced feedback has received less attention, although it is known that explicit visual feedback on balance decreases postural sway in healthy and patient populations [210,211]. Potentially, the availability of such explicit visual information leads to down-weighting of other, less unambiguous, sensory information. This would have clinical relevance as explicit visual feedback on balance is currently used in diagnostic tests and training and rehabilitation programs for balance control [212]. If down-weighting of other sensory modalities occurs, impairments in these systems might be overlooked in diagnostic tests and training may be sub-optimal. Despite the fact that many studies have explored the effects of sensory manipulations and visual feedback on balance control, to our knowledge, none has explored their interaction [212,213]. Therefore, a better understanding of the interaction of visual feedback and sensory impairments is necessary.

Mediolateral balance is of particular interest, since it is more affected by ageing and disease and since its deterioration has been associated with an increased risk of falling [151,156,169,181,214]. Therefore, this study will focus on balance performance in the frontal plane, and the effects of non-explicit (normal visual information) and explicit feedback on centre of mass (CoM) sway under sensory manipulation conditions.

The aim of this study was to determine whether the effect of vestibular, proprioceptive and somatosensory manipulations is reduced by the presence of visual information and by explicit feedback on body CoM sway. We hypothesized that manipulation of sensory systems increases mediolateral CoM sway but that this effect is reduced by presence of visual information and even more by explicit feedback on CoM sway.

Methods

Subjects

Nineteen healthy young adults, 12 women and 7 men, participated in this study (age: 28±3 years; height: 1.75±0.10 m; weight: 70.0±8.0 kg). Participants did not report any musculoskeletal or neurological conditions that could affect balance. This research was approved by the local Ethical Committee, in accordance with the ethical standards of the Declaration of Helsinki. All subjects were informed of the experimental procedures and signed an informed consent form prior to the experiment.

Task and Procedure

Each participant performed a series of 50 seconds standing tasks while barefoot and with the arms crossed on the chest. CoM data were obtained using a 9 markers, 2D kinematic model.

Markers were located at the forehead, mid shoulders, anterior-superior iliac spines (ASIS), mid knees and mid ankles and captured with an Optotrak Certus motion capture system (Northern Digital Instruments, Canada). Gender specific CoM calculations were performed using scaling of anthropometric data and of inertial parameters described by de Leva [215]. For set-up and model details refer to figure 2.1. D-flow 3.10.0 software (Motek Medical, The Netherlands) was used to produce a stationary target (11 cm diameter white sphere) in the middle of a screen located 2.5 m in front of the participants, as well as to display and record CoM ML displacements (9 cm diameter red sphere). The delay of the system was calculated to be 16 ms which is equivalent to 1 sample.

Displacement of the CoM in the frontal plane when standing still was measured under three conditions: eyes open with explicit visual feedback (FB) on ML body CoM sway, eyes open (EO) without explicit feedback and eyes closed (EC). Each of these conditions was measured three times

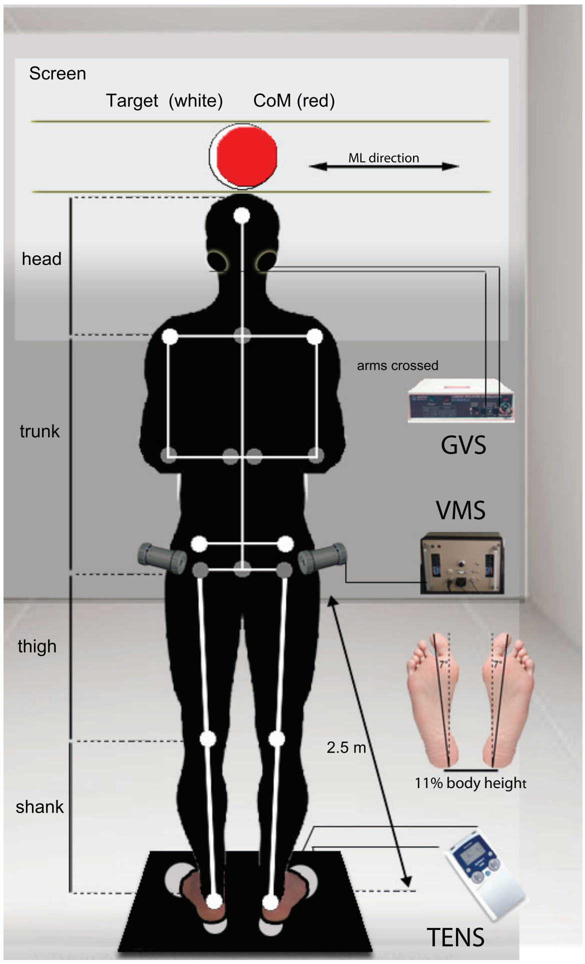

Figure 2.1. Illustration of the set-up and the model for CoM calculation utilized in this experiment, showing a silhouette of a subject with marker placement superimposed (in white actual makers and in grey estimated joint centres) and the display of the CoM feedback (red sphere). The white sphere in the centre represents a static target placed in the middle of the screen. The allocation of GVS electrodes (mastoid processes), VMS vibrators (half way between iliac crests and major trochanters) and TENS electrodes (grey areas below the feet) are also depicted. An insertion of foot soles is presented showing foot positioning during` the experiments (stance width and angle).

for 50 seconds. These conditions were also measured under three types of sensory manipulations: somatosensory, vestibular and proprioceptive. For the somatosensory manipulation, we used a transcutaneous electric nerve stimulator (TENS, ElphaII 3000, Danmeter, Denmark) to apply a current to the feet soles through an aluminum plate under the feet. The stimulus was delivered with the current flow alternating from left to right at 0.25 Hz (ramp up: 0.5 s; plateau: 1 s; ramp down: 0.5 s) with a maximum of 18 mA intensity at 120 Hz. For the vestibular manipulation, we used a linear isolated stimulator (Stimsola, Biopac systems, Inc., USA) to produce galvanic vestibular stimulation (GVS) with a maximum intensity of 1.5 mA. Flexible carbon electrodes (5x5 cm) were allocated on the mastoid region using conductive gel and tape for fixation. Finally, for the proprioceptive manipulation, we used vibrational muscle-tendon stimulation (VMS) and attached a pair of custom-made vibrators (100 Hz vibration) bilaterally over the gluteus-medius topographic region. These muscles were selected since they are the main actuators involved in maintaining upright stance in the frontal plane [208]. For both GVS and VMS, we constructed a left/right alternating frequency as the sum of 6 sine waves (0.15, 0.25, 0.35, 0.45, 0.55 and 0.65 Hz). For the GVS, all signal peaks were normalized to 1.5 mA whereas for the VMS, peaks were squared and normalized to 4 V (equivalent to 100 Hz). Due to these normalizations, the main frequency content was 0.45, 0.55 and 0.65 Hz for both GVS and VMS signals (figure 2.2). In total each participant performed 36 trials (3 conditions x 4 manipulations x 3 trials). Subjects were given at least 1 minute rest in between trials not reporting fatigue at the end of the session. A Borg scale was used to quantify the perceived exertion, when reaching a score of 4 (sort of hard) subjects were enforced to take longer rests periods until perceived exertion reached below 3 (moderate).

Data Analysis

The ML CoM sway power spectrum was calculated using a 20*sample rate (60 Hz) discrete Fourier transform (FFT) with zero-padding as required to obtain 0.05 Hz resolution. The sum of all values of energy in the spectrum analyzed ($E_{spectrum}$) was used as a measure of the overall CoM sway. A separate analysis on the frequency response at those frequencies present (E_{input}) and those not present (E_{non}) in the sensory manipulations was performed. For the GVS and VMS manipulations the sum of energy values from 0.05 to 1.95 Hz at steps of 0.1 were used to determine E_{input} whereas for the TENS these were .25, .75, 1.25 and 1.75 Hz. The remaining energy values for each manipulation were used to calculate $E_{non.}$

Statistical Analysis

Three repeated measures 2-way ANOVAs were used to determine significant differences in ML CoM sway for the variables $E_{spectrum}$, E_{input}, and E_{non} between each modality of sensory manipulation (TENS, GVS VMS) and the control condition without any sensory manipulation and between visual input conditions (FB, EO and EC), as well as the interaction effects. Paired comparisons were used to test for differences between each of the visual feedback conditions. Descriptive statistics were performed to determine the direction of the differences. A Bonferroni adjustment was made to the α-level; to achieve significance at 0.05, the *p*-value had to be below 0.0167 (0.05/3).

Results

Figure 2.2 shows a typical example of the ML sway of one subject under each of the experimental conditions. As can be seen, ML CoM sway is reduced when providing visual feedback (FB) but no differences can be observed between EO and EC conditions. The shape of the sensory manipulation signals (TENS, GVS and VMS) is also shown. The positive values in this figure indicate right side stimulation and CoM displacement whereas negative values indicate the same

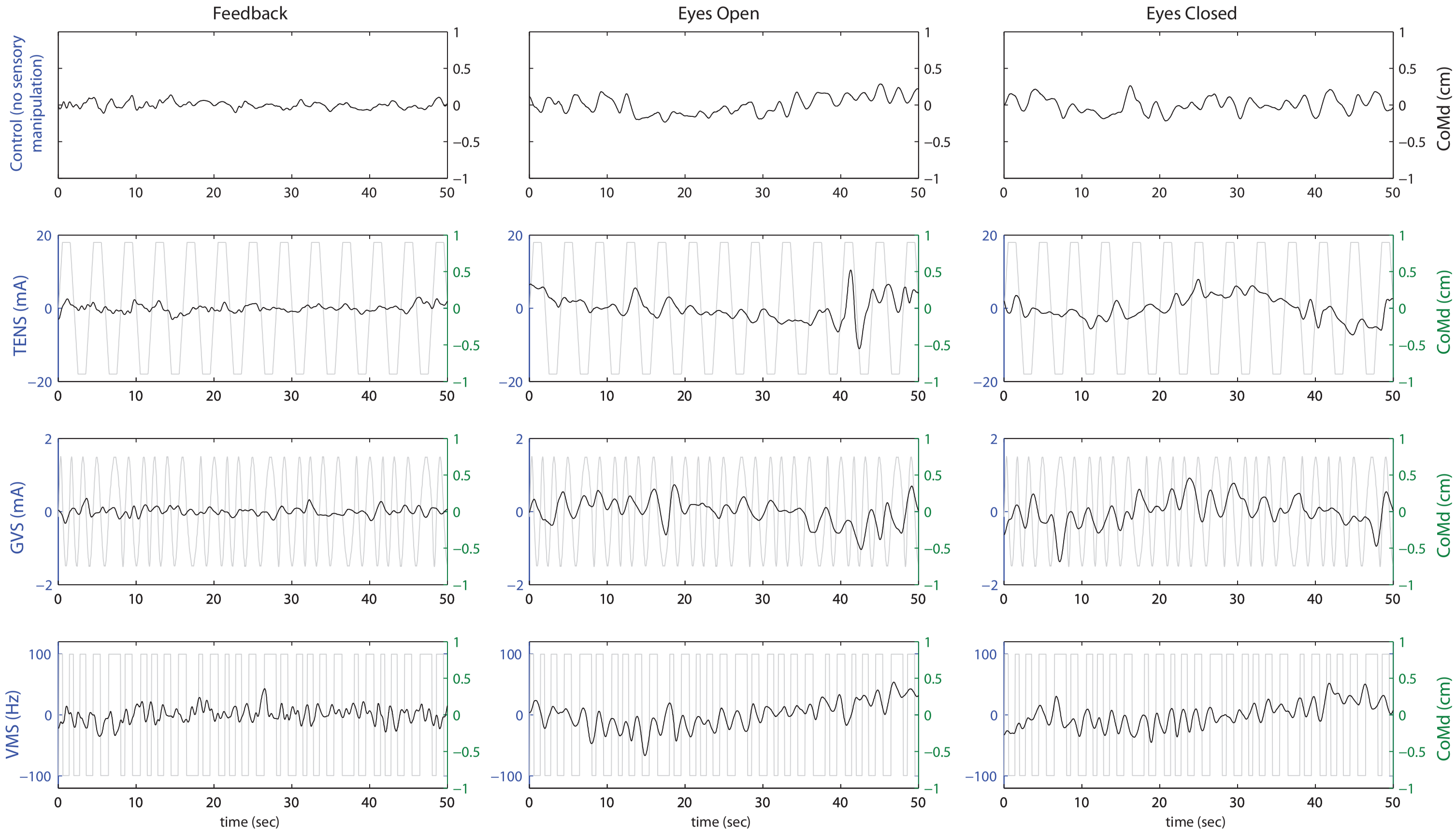

Figure 2.2. From top to bottom plots represent the CoM ML sway (black lines) of a representative subject in the control condition and with TENS, GVS and VMS manipulations and, from left to right, in feedback, eyes open, eyes closed conditions. In grey the shapes of the TENS, GVS and VMS signals are presented. The positive values in this figure indicate right side stimulation (left axis) and CoM displacement (right axis) whereas negative values indicate the same but to the left.

but to the left. Averaged CoM sway for the control condition, TENS, GVS and VMS at the three visual input conditions (FB, EO and EC) is presented in figure 2.3, whereas the means for $E_{spectrum}$, E_{input}, and E_{non} across all conditions are presented in figure 2.4. Table 2.1 presents the means and results of statistical tests for all the variables measured. Overall, sensory manipulations increased the means of all variables analysed when compared to control conditions for each of the visual input conditions. Also, FB conditions exhibited the lowest values for all variables analyzed when compared to EO and EC across all sensory manipulations yet no differences were found between EO and EC conditions.

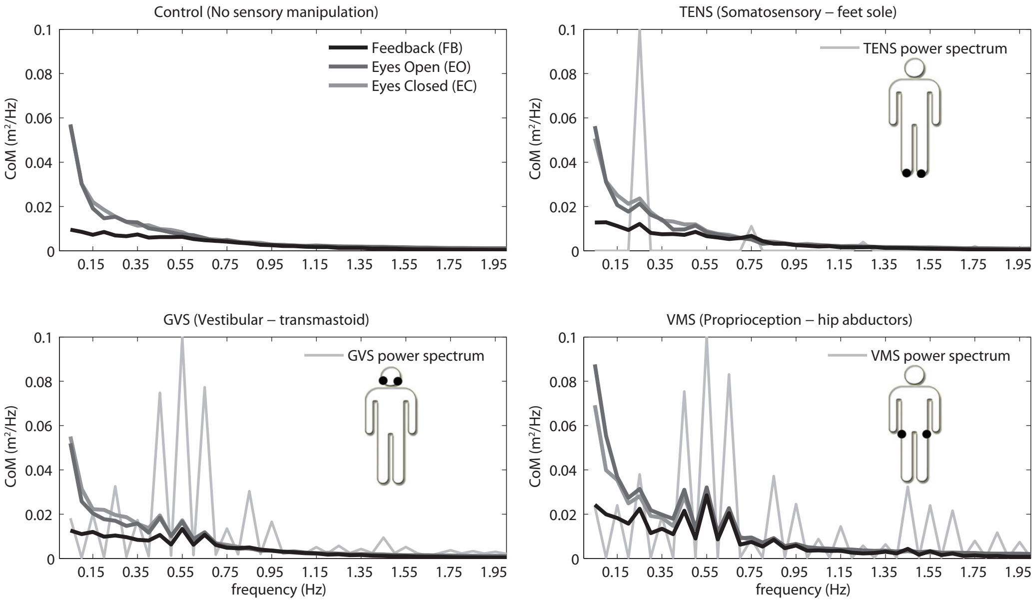

Figure 2.3. The averaged ML CoM power spectrum curves under the three visual conditions in separate plots for each of the sensory manipulations (Control, TENS, GVS and VMS). The abscise shows absolute values of energy calculated for the CoM-ML displacement for each of the frequencies in the .05 to 2.0 Hz range. Shaded grey areas represent ± sd. The lightest grey curve in the TENS, GVS and VMS plots shows the power spectrum curve for the sensory manipulations signal.

Manipulation of sensory information from the feet soles (TENS).

When compared to the control condition, the effect of TENS on sway was only significant at the input frequency (E_{input}). Significant effects of visual input were also found, with the lowest values of $E_{spectrum}$, E_{input}, and E_{non} observed in the feedback condition. No interaction effects of TENS and visual conditions were found for any of the variables.

Manipulation of vestibular information (GVS).

In comparison to the control condition, GVS significantly increased sway across the whole power spectrum analyzed ($E_{spectrum}$) as well as for the E_{input}, and E_{non} measures. Also significant effects of visual input were found for these three variables, again, with the lowest values in the feedback condition. No interaction effects of GVS and visual conditions were found for any of the variables.

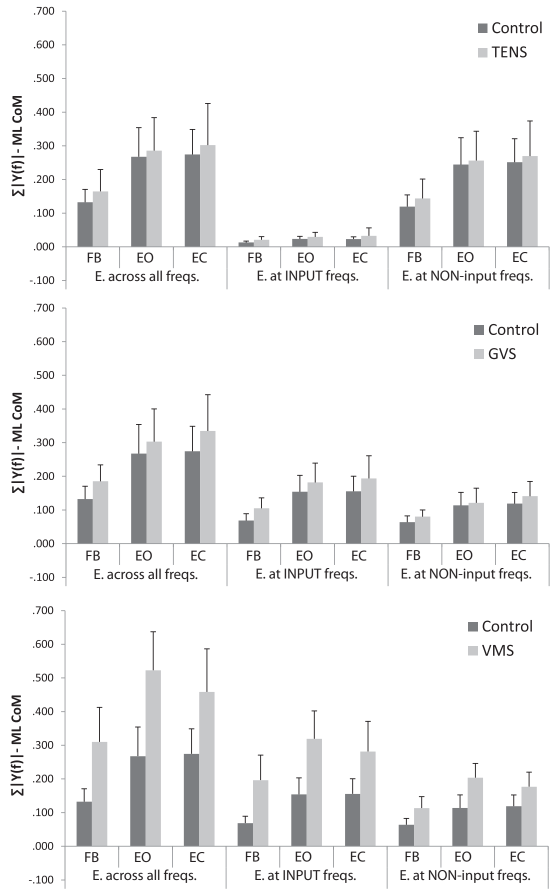

Figure 2.4. Light grey bars represent the means ±sd of $E_{spectrum}$, E_{input}, and E_{non} when applying TENS (top left), GVS (bottom left) and VMS (top right) manipulations and the dark bars show the same variables in the control condition. Conditions: FB = feedback; EO = eyes open; EC = eyes closed.

Table 2.1. Descriptive statistics (mean±sd) for Espectrum, Einput and Enon across all visual input conditions (feedback, eyes open and closed) and sensory manipulations. Right side of the table shows p-values for the RM-ANOVAs using manipulation (TENS, GVS or VMS) and visual input (eyes open and closed) as main factors and their interactions.

		Feedback		Eyes Open		Eyes Closed				
		mean	sd	mean	sd	mean	sd	**Manipulation**	**Visual input**	**Manipulation * Visual input**
Control	$E_{spectrum}$	.132	.038	.268	.086	.274	.074			
	E_{input} [a]	.069	.020	.154	.049	.156	.045			
	E_{input} [b]	.013	.004	.023	.008	.023	.007			
	E_{non} [a]	.064	.019	.114	.039	.119	.035			
	E_{non} [b]	.119	.035	.244	.080	.251	.035			
TENS	$E_{spectrum}$	.165	.065	.286	.098	.302	.124	.121	**<.001**	.798
	E_{input}	.021	.009	.029	.014	.033	.024	**.005**	**<.001**	.766
	E_{non}	.144	.058	.256	.087	.270	.104	.205	**<.001**	.834
GVS	$E_{spectrum}$	.185	.049	.303	.097	.335	.108	**.002**	**<.001**	.658
	E_{input}	.105	.031	.182	.058	.194	.067	**.001**	**<.001**	.827
	E_{non}	.080	.020	.121	.043	.141	.044	**.016**	**<.001**	.463
VMS	$E_{spectrum}$	.310	.103	.523	.114	.458	.128	**<.001**	**<.001**	.046
	E_{input}	.196	.074	.319	.083	.281	.089	**<.001**	**<.001**	.149
	E_{non}	.113	.034	.204	.042	.177	.043	**<.001**	**<.001**	**.011**

[a] Sum of values at TENS input frequencies and non-input frequencies, respectively (input frequencies = .25, .75, 1.25 and 1.75 Hz).
[b] Sum of values at GVS/VMS input frequencies and non-input frequencies, respectively (input frequencies = .05:0.1: 1.95 Hz).

Manipulation of proprioceptive information (VMS).

A significant effect of VMS was found across all variables analyzed when compared to the control condition. Significantly lower values for $E_{spectrum}$, E_{input}, and E_{non} were also observed when visual feedback was available. A significant interaction effect was found for E_{non}. Although not significant, a similar interaction trend was also observed for $E_{spectrum}$ (p = 0.046). However, these interactions indicated a stronger effect of VMS with EO than with EC, which was not in line with the hypothesized effect of visual feedback.

Discussion

This study aimed to determine whether the presence of visual information and explicit feedback of body CoM sway reduce the effect of somatosensory, vestibular and proprioceptive manipulations on CoM sway. We showed that somatosensory, vestibular and proprioceptive manipulations lead to increased body CoM sway at the frequencies of the input signals. Absence of normal visual information had no effect on ML CoM sway, while explicit feedback on body CoM reduced ML sway, but did not change the effects of sensory manipulations on ML sway.

Sensory manipulations

To our knowledge, this was the first study using TENS to disturb somatosensory inputs from the foot sole receptors during standing. The main effect of TENS is the depolarization of Aβ (II) somatosensory fibers, present in pressure receptors in the foot sole [15,216]. Application of TENS bilaterally to the posterior aspect of the leg and below the sensory threshold has been shown to reduce postural sway [216]. Since we switched the TENS signal side-to-side at 0.25 Hz with a fixed, supra-threshold 18 mA intensity, we expected an opposite effect. This effect was reflected in E_{input} for which a significant increase with TENS was found. Nevertheless, no significant effects were found on $E_{spectrum}$ and E_{non}, which indicates an overall limited and quite specific effect of somatosensory input from the foot sole on body CoM sway. However, no significant findings can be also explained by the lower complexity of the TENS stimulation signal (compared to GVS and VMS) for which balance control responds in a more linear fashion. Single tone somatosensory stimulation may have not demanded central processing (sensory reweighting) but rather reflexive responses may have been elicited. Note that for none of the subjects the selected intensity produced any visible or reported muscle contraction, which could have caused CoM sway towards the side contralateral to the stimulated foot. Although CoM sway was lower with feedback than in eyes open and closed conditions, feedback did not overcome the effects of TENS.

GVS stimulates the otholith organs as well as the semicircular canals, which elicits balance responses that incline the body towards the anodal side [20]. In our setup, the anodal side alternated from left to right, which induced sway at the frequencies contained in the GVS signal (E_{input}), as was previously shown for sway of the head [217]. With regards to the effect of visual inputs, the feedback condition exhibited the lowest energy values for all the variables measured. Yet, even with explicit feedback, sway was still significantly larger with GVS than in the control condition. Possibly, responses arising from vestibulospinal reflexes may account for this difference, as it has been previously shown that making sensory channels available (i.e. opening the eyes) while perturbing vestibular inputs does not completely remove short and medium latencies vestibulospinal reflexes [31].

VMS predominantly activates type Ia afferents of muscle spindles, which may cause reflexive muscle activation and in addition the subject perceives “lengthening” of the muscle, which may cause voluntary activation, leading to responses that increase sway at the input frequency [218,219]. Invalid input may cause a decreased use (down-weighting) of the Ia-afference input. Under the feedback condition, enhanced (vision) and non-perturbed (vestibular and somatosensory) sensory channels could theoretically compensate for the perturbed Ia-afference. However, VMS resulted in larger ML sway for all visual input conditions compared to the control condition, which may reflect an inability to generate inhibitory mechanisms for reflexive responses and/or down-weighting of illusory muscle lengthening. Significantly lower energy values were observed for feedback compared to eyes open and closed conditions, nonetheless, an interaction of vision with VMS was observed only for E_{non}. Although not significant for $E_{spectrum}$ and E_{input}, less sway in EC compared to EO conditions may reflect a tighter balance control called into play when proprioception is disturbed in the absence of visual inputs. A possible strategy to reach such tight control is by increasing stiffness through increased muscle co-activation of the main ML stabilizers at both sides. This is a strategy observed in the elderly while standing and during gait [220,221].

For both GVS and VMS, significant main effects at E_{non} indicate non-linear sway responses possibly arising from the interaction of reflexive and corrective balance responses. The latter may be in response to the immediate perturbation of the upright posture due to GVS-induced paraspinal reflexes or VMS-induced stretch reflexes. However, illusory sensations of balance disturbance (muscle lengthening and whole body inclination) can also elicit these corrective responses. Since GVS and VMS manipulations alternated from left to right, corrective responses may occur at a phase lag with respect to the reflexes, which would contribute to the main effects on $E_{non.}$

It has been reported that not all individuals are able to take advantage of visual feedback to enhance balance control [222] and that, in order to reduce postural sway, visual CoP feedback must have gain of at least 5 on display [223]. In our experiment, the gain for visual display of the CoP movement ranged from 7 to 9 between subjects, due to normalization for height, which ensured noticeable movement of CoM on the display. It is noteworthy that all subjects in this experiment decreased ML CoM sway in the frequency range analysed when visual feedback was presented.

Visual information

As shown in previous studies, explicit visual feedback reduced CoM sway when compared to eyes closed conditions [210,211,222]. However, no reduction in sway was found in the EO compared to the EC condition. Similar results were observed by Hay et. al. who found that young adults are able This suggests a limited importance of visual information for balance control under normal conditions, but highlights the possibility of increasing its relative contribution, as demonstrated by a significant reduction in energy across the whole power spectrum analysed when explicit feedback was provided.

Evidence suggests that young adults rely less on vision than older adults [90,224]. In line with our findings , in young adults, vision accounts for 10% of sensory inputs used for balance control when standing [21]. A larger contribution of vision could be expected during application of the sensory manipulations. However, differences between EO and EC conditions were not significant during TENS and GVS and actually opposite to expectations during VMS. This study was conducted in healthy young adults for which indemnity of all sensory systems as well as weighting abilities is expected. Therefore, for the GVS condition with EC, proprioceptive and somatosensory inputs are likely to be up weighted whereas for VMS with EC, vestibular, somatosensory and non-manipulated proprioceptive (i.e. ankle muscles) afferents may compensate the absence of visual inputs. *Limitations*

This study had some limitations. Firstly, the TENS device used did not allow to fully manipulate stimulation parameters, so as to create similar frequency content signals as in GVS and VMS. Secondly, the VMS signal was built to instantaneously reach 100 Hz, to exceed proprioceptors thresholds, this differs from the sinusoidal GVS or ramped (up and down) TENS stimulations signals. Finally, the intensity of the different sensory manipulation modalities was not scaled. Due to these factors quantitative comparisons of the somatosensory, proprioceptive, and vestibular manipulations cannot be made.

Practical implications

Although the use of visual feedback on balance performance has previously been used to assess balance [225,226], its interaction with disturbances of sensory systems as a model for sensory impairment had not yet been studied. Since upweighting of visual information might occur when explicit visual feedback is provided, diagnostic tests may overlook impairments of other sensory systems contributing to balance control. However, our results showed no interactions between visual information and sensory manipulations of the somatosensory, vestibular and proprioceptive systems. The results of this investigation also show that ML sway was significantly increased for all manipulation modalities at their respective frequency content compared to control conditions even when explicit feedback was provided. Although effects of sensory manipulations will differ from the physiological changes in acuity of the senses with aging or due to pathology, it is likely that the use of feedback (up-weighting of visual inputs) on balance assessment cannot overcome and therefore mask sensory deficits.

Conclusion

Electrical stimulation with current through the soles of the feet and the vestibular system, alternating from left to right, as well as alternating vibratory stimulation of the muscle spindles in the hip abductor muscles, lead to increased sway at the frequencies of the input signals. Absence of normal visual information had no effect on ML CoM sway, while explicit feedback on CoM reduced ML sway, although it did not efface the effect of manipulations of the somatosensory, vestibular and proprioceptive systems. Non-linear effects of vestibular and proprioceptive disturbances occurred, which may indicate an interaction of reflexive and voluntary responses in balance control. Since no confounding effect of up-weighting of visual inputs due to explicit feedback was found, this investigation highlights the potential of using feedback to assess and train balance.

III

Frequency domain mediolateral balance assessment using a centre of pressure tracking task

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Abstract

Since impaired mediolateral balance can increase fall risk, especially in the elderly, its quantification and training might be a powerful preventive tool. We propose a visual tracking task (VTT) with increasing frequencies (0.3-2.0 Hz) and with centre of pressure as visual feedback as an assessment method. This mediolateral balance assessment (MELBA) consists of two tasks, tracking a predictable target signal to determine physical capacity and tracking an unpredictable target signal to determine sensorimotor integration limitations. Within and between sessions learning effects and reliability in balance performance descriptors in both tasks were studied in 20 young adults. Balance performance was expressed as the phase-shift (PS) and gain (G) between the target and CoP in the frequency domain and cut-off frequencies at which the performance dropped. Results showed significant differences between the MELBA tasks in PS and G indicating a lower delay and higher accuracy in tracking the predictable target. Significant within and between sessions learning effects for the same measures were found only for the unpredictable task. Reliability of the cut-off frequencies at which PS and G performance declined and the average values within cut-off frequencies was fair to good (ICC 0.46-0.66) for the unpredictable task and fair to excellent for the predictable task (ICC 0.68-0.87). In conclusion, MELBA can reliably quantify balance performance using a predictable VTT. Additionally, the unpredictable tasks can give insight into the visuomotor integration mechanisms controlling balance and highlights MELBA's potential as a training tool.

Introduction

Balance impairments are a common cause of falls in the elderly population [5,151,227]. Detriments of the somatosensory and neuromuscular systems have been identified as causes of imbalance when standing and walking in the elderly [74,78,138,172,227-230]. The inability to adequately integrate sensory inputs as well as difficulties to perform dual-tasks in which cognition is required have also been identified as causes of balance impairments in older adults [18,90,231,232]. Previous investigations have demonstrated that several biomechanical variables of balance control in the mediolateral (ML) direction can identify fallers when standing (i.e. ML postural sway measures) and when inducing sideward stepping responses [151,156,169,172,181,233]. There are also indications that centre of mass displacement (CoM) in the frontal plane, when compared to sagittal, requires greater active control when walking [234-236]. Furthermore, evidence has shown that balance training targeting movements in the frontal plane may reduce the incidence of falls in community-dwelling elderly people [237-239].

Despite the discriminative capacity (fallers from non-fallers) of ML balance control reported in retrospective studies, only two of the biomechanical variables (i.e., spontaneous sway of the centre of pressure (CoP) during quiet standing and gluteus medius onset time in a stepping response task) have shown poor to moderate accuracy in prospectively predicting falls [151,156]. It is possible that due to a ceiling effect of current balance assessment tools, including clinical measures, those tests are not sensitive enough to detect balance impairments and predict falls in high functioning elderly and in able-bodied subjects [5,156,240-242]. Therefore, a more sensitive assessment method should challenge balance more to avoid ceiling effects and yet be simple enough to be applied in a clinical environment [242,243]. In this context, we propose a mediolateral balance assessment (MELBA) method, which uses a visual tracking task (VTT) and the ML CoP displacement as feedback on performance.

In the VTT subjects have to elicit voluntary ML CoP movements based on visual information of the target and subordinating proprioceptive and vestibular sub-systems to maintain stability. By increasing task difficulty (i.e. increasing target frequencies), the subject is challenged to respond fast and accurately. These aspects of the response are necessary when coping with perturbations in daily life situations and reflect the integrity and compensatory ability of the balance system. MELBA aims to quantify balance performance using a visual tracking task, whereby balance control is then characterized by gain and phase-shift between target and CoP signals. MELBA consists of a predictable target, which allows feed-forward mechanisms to control balance in order to determine maximal physical capacities, and a second, unpredictable target, which increases the demand of feedback mechanisms in order to quantify limitations in sensory integration.

This study aimed to determine the methodological properties of MELBA by assessing learning effects within and between sessions as well as reliability of the performance, i.e. the consistency of the method when no interventions are made. Additionally, balance performance between the two MELBA tasks (i.e., predictable versus unpredictable) was compared.

Methods

Subjects

Twenty healthy young adults, 12 women and 8 men, participated in this study (age: 28±3 years; height: 1.75±0.1 m; weight: 70±8 kg). Participants did not report any musculoskeletal or neurological condition that may have affected balance. This research was approved by the local Ethical Committee, in accordance with the ethical standards of the declaration of Helsinki. All subjects were informed of the experimental procedures and signed an informed consent form prior to the experiment.

Task and Procedure

Each participant performed a series of visual tracking tasks (VTT) while standing barefoot and with the arms crossed on a force-plate located in a quiet and low-intensity light room (for set-up details see figure 3.1). CoP data were obtained using a Kistler-9281B force plate (Kistler Instruments AG, Winterthur, Switzerland) sampling at 60 samples/s. D-flow 3.10.0 software (Motek Medical, The Netherlands) was used to produce target signals as well as to record and display target and CoP data on the screen. The delay of the system was calculated to be 16 ms which is equivalent to 1 sample.

A *predictable* target signal was constructed using 18 blocks of 5 seconds, each composed by one sine wave, which increased from 0.3 to 2.0 Hz in steps of 0.1 Hz. This information was enhanced using a metronome synchronized with the maximum displacement of the target in order to increase sensory input abundance. The total task time was 90 seconds.

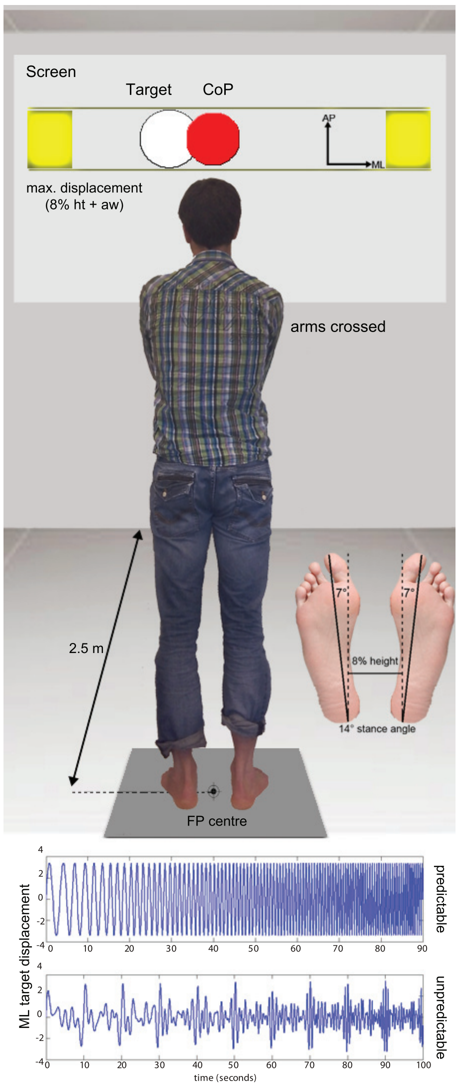

Figure 3.1. Illustration of the experimental set up from a rear view. The CoP (centre of pressure) is represented as red sphere, whereas a white sphere represents the target signal, on a display located 2.5 m in front of the forceplate (FP). The height (ht) and ankle width (aw) were used to determine the maximum amplitude of the target indicated by the yellow areas projected at both sides of a corridor delimited by a top and bottom horizontal bars. These bars indicate the tolerance for anterior-posterior (AP) displacement which was set at 2 cm. Additional auditory feedback ("beep") was given when these AP boundaries were exceeded. The figure of feel soles inserted at the right depicts foot positioning across all trials (7°rotation of each foot with an intermalleoli distance equal to 8% of ht). The lower panel depicts the predictable (top) and unpredictable (bottom) targets. Negative and positive values indicate left and right CoP displacements, respectively.

An *unpredictable* target signal was constructed using 15 blocks composed by the sum of 6 consecutive sine waves separated by 0.1 Hz. A pseudorandom phase-shift between sine waves between -1 to 1 periods was introduced in order to avoid predictability. After each block the lowest frequency, which started at 0.1 Hz was increased by 0.1 Hz higher until it reached 1.5 Hz. Duration was 10 s for blocks 1 and 2, 8 s for blocks 3 to 7, 6s for blocks 8 to 11 and 4 s for blocks 12 to 15. Duration of the blocks was chosen in order to obtain at minimum of 2 cycles per frequency contained in the block. This target construction also allowed limiting the total task time to 100 seconds. The unpredictable target bandwidth started at a lower frequency than the predictable target, but results in the frequency range 0.1-0.2 Hz were not analyzed. An example of the two target signals is depicted in figure 3.1.

The VTT consisted of tracking the predictable or unpredictable target signal using the ML displacement of the CoP projected on a screen in front of the subject. The screen (2 x 1.5 m size) was placed 2.5 m in front of the force plate centre. The target signal and CoP were represented by white and red spheres of 11 and 9 cm diameter, respectively. Each participant performed 8 VTT trials: 4 with the predictable and 4 with the unpredictable target. Before performing the tasks two practice trials were allowed for each of the conditions. This session was repeated 7 days later at the same time of the day in order to avoid the influence of the circadian cycle in the variables measured [244]. Trials were performed with at least with 1 minute rest in between. Since stance width alters lower limb neuromechanical responses when displacing CoM and CoP in the ML direction [208,245], stance width was standardized by setting the intermalleoli distance to 8% of body height. A fixed 14° stance angle was used across all participants (figure 3.1). These stance measures have been shown to be within the values of normal stance [246]. Target maximum side-to-side displacement for both target signals was normalized for each subject at stance width plus ankle width; allowing CoP ML displacements to be within the BoS. On average, participants stood on the force plate with 14±0.8 cm distance between malleolus or stance width, being 7.1±0.6 cm the average ankle width which determined a maximum target displacement of 21.1±1.3 cm. The projected maximum target displacement was 1.4 m side-to-side, therefore the gain factor for CoP displacements projected on the screen ranged from 6.0 to 7.5 with an average of 6.7 ± 0.4. Anterior-posterior displacements of the CoP were constrained by allowing only 3.2 cm fore-aft displacement around the CoP at rest, which was indicated by the projection of two yellow bars above and below the target signal. The computer produced an auditory cue whenever CoP movement exceeded these constraints.

Data analysis

All data analysis was performed using custom-made software in Matlab R2011a (Mathworks, Natick MA, USA). Balance performance over the frequency ranges in the target signal was described by gain, and phase–shift of the linear constant coefficient transfer function between CoP and target signal. This analysis was performed using the Welch algorithm over windows of 0.25 times the length of the target (per block) with 90% overlap between windows and frequency resolution of 0.1 Hz. For each block we maintained estimates of gain and phase-shift only for the frequencies present in the target presented in that block. The results for all blocks of one trial were combined after estimation of the transfer functions, to obtain gain and phase-shift values at all frequencies presented. For the unpredictable target, phase-shift, gain and coherence were calculated as the average of the values at each frequency over blocks with overlapping frequency content. The phase-shift (PS) reflects the delay (in degrees) between target and CoP at each frequency, whereas gain (G) reflects the ratio between the target and CoP amplitudes; both in the frequency domain. Perfect performance implies PS = 0 and G = 1. In addition, the coherence (Coh) was determined, as a measure of linear correlation between the target and CoP in the frequency domain, which in this study was used to corroborate the assumption of input (target)/output (CoP) linearity and therewith the validity of estimates of PS and G. Considering that each coherence estimate was determined from 16 independent data windows (4 windows per block times 4 trials), the threshold for significance of coherence can be estimated at 0.181 [247,248] To characterize balance performance, 4 descriptors were calculated. First, the average of the three highest values for each measure in each trial was calculated and the values at which PS and G dropped below 75% of this average were determined as the cut-off frequencies (coined PSX and GX). Second, PSY and GY were computed as the average of the G and PS values within the bandwidth determined by PSX and GX, respectively. For better clarity, calculations of performance descriptors are illustrated in figure 3.2.

Statistical Analysis

To test for learning effects, repeated measures ANOVAs were performed on the average of phase-shift and gain over all frequencies and for the dependent measures PSX, GX, PSY and GY using 2 (predictable and unpredictable target) x 2 (assessment day) x 4 (trial number) models. In

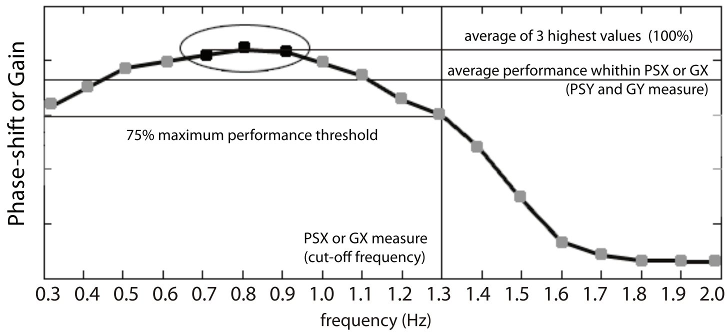

Figure 3.2. Figure illustrates the calculation of performance descriptors. First, the three highest values for G and PS were indentified (circled black dots) and averaged. Second, 75% of the previously calculated mean was used as performance threshold for G and PS. The frequency (Hz) of last values above this threshold was used as cutoff frequency for PSX and GX, respectively (vertical dashed line). Finally, the averages of the PS and G values within PSX and GX, (PSY and GY, respectively) characterized performance within the cutoff frequency (horizontal line).

view of multiple testing, α was Bonferroni corrected at 0.0083 (0.05/6). To determine between-days reliability of performance descriptors, further analysis included intraclass correlations (ICC 2,1) for absolute agreement by using descriptors averaged within assessment days. Measures were considered to exhibit excellent reliability when ICC > 0.75 and fair to good reliability when ICC value was between 0.4 and 0.75 [249]. Assumption of normality of the data was confirmed by Shapiro-Wilks tests. Statistical analyses were performed using SPSS (PASW statistics 18) and Matlab.

Results

Participants did not report fatigue during or after the trials and were able to complete all trials. Figure 3.3 illustrates average performance over subjects when tracking both targets. Descriptive statistics and repeated measures ANOVAs for the measures of PS and G are summarized in table 3.1.

Average values for Coh for the predictable target were .88 and .89, whereas for the unpredictable target these were .45 and .52 for sessions 1 and 2, respectively. These values were high above chance levels indications that the relation between the CoP and the predictable target was sufficiently strong and linear to allow estimation of transfer functions between target and CoP signals.

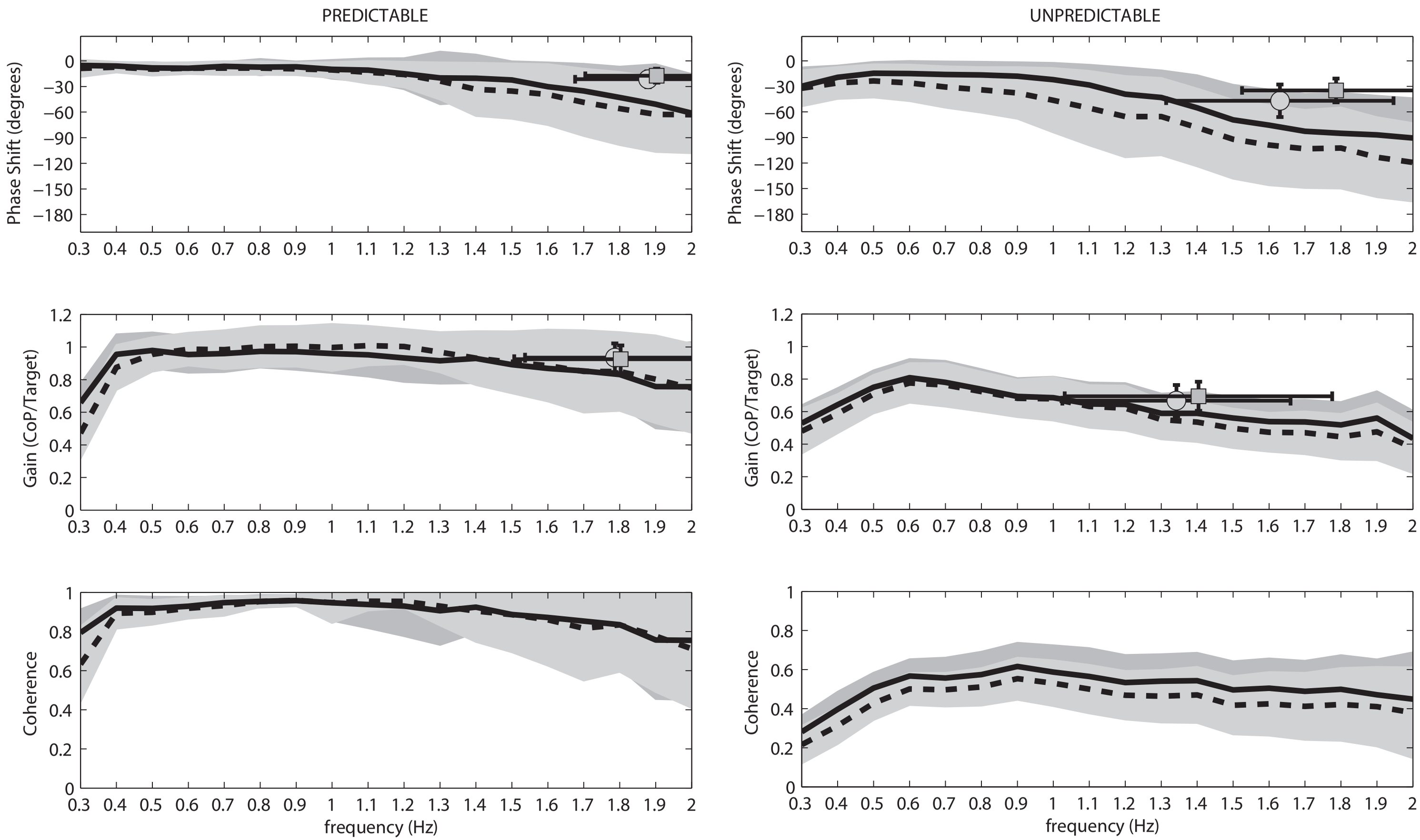

Figure 3.3. Averaged curves (± sd) for phase shift (top panel), gain (mid panel) and coherence (bottom panel) measures for the predictable target (left) and unpredictable (right) targets, during first (dashed line) and second (continuous line) sessions. Grey shading indicates the ± sd for all subjects and for all trials. Crosses inserted in the plots indicate means (± sd) for performance descriptors for the first session (circular markers) and second session (square markers).

Table 3.1. Descriptive statistics for phase shift, gain and coherence and performance descriptors (PSX, PSY, GX and GY) for the predictable and unpredictable targets at both assessment days. Right part of the table summarizes the p-values of the repeated measures ANOVAs for the main factors target (unpredictable and predictable), day (initial assessment and assessment seven days later), and trial (trials 1 to 4) and all interactions effects. Statistically significant p-values are presented in bold.

	PREDICTABLE				**UNPREDICTABLE**				***p* -values**						
	Day 1		**Day 2**		**Day 1**		**Day 2**								
	mean	**sd**	**mean**	**sd**	**mean**	**sd**	**mean**	**sd**	Target	Day	Trial	Target x Day	Target x Trial	Day x Trial	Target x Day x Trial
Phase Shift	-0.39	0.32	-0.32	0.27	-1.04	0.56	-0.7	0.46	**<.001**	**<.001**	**<.001**	**<.001**	**<.001**	0.352	0.549
PSX (Hz)	1.89	0.18	1.9	0.22	1.67	0.32	1.82	0.25	**<.001**	**0.004**	0.706	0.013	0.205	0.777	0.866
PSY (°)	-0.33	-0.15	-0.26	-0.14	-0.77	0.34	-0.55	0.24	**<.001**	**<.001**	0.32	**0.004**	0.617	0.895	0.926
Gain	0.9	0.13	0.89	0.09	0.59	0.12	0.63	0.1	**<.001**	**<.001**	**<.001**	**<.001**	0.055	**0.009**	0.054
GX (Hz)	1.79	0.25	1.8	0.3	1.34	0.32	1.41	0.37	**<.001**	0.088	0.313	0.19	0.327	0.945	0.999
GY	0.93	0.09	0.93	0.08	0.67	0.1	0.7	0.09	**<.001**	0.48	0.692	0.97	0.955	0.812	0.822
Coherence	0.88	0.09	0.89	0.07	0.45	0.08	0.52	0.08	**<.001**	**<.001**	**<.001**	**<.001**	**<.001**	0.023	0.395

Overall, a significant main effect was found for target (Table 3.1), indicating that participants exhibited better mediolateral balance performance (PS, G) when tracking the predictable target compared to the unpredictable target (Fig 3.2). On average, when tracking the predictable target, participants performed nearly in-phase (PS ≥ 0) and close to optimal (G ≥ 1) for input frequencies below 1.2 Hz. For the unpredictable target, near-optimal performance values for PS and G were only observed during the second session in between 0.4 to 1.0 Hz.

Cut-off frequencies for PS and G (PSX and GX, respectively) were significantly higher when tracking the predictable target compared to the unpredictable target as indicated by a main effect of target on these parameters (table 3.1 and fig 3.2). A similar effect was observed for PSY (greater PS) and GY (higher G). The between sessions effect was significant for PSX and PSY showing a learning effect after one week, whereas this learning effect was not statistically significant for GX and GY. A main effect of trial was observed only for PSY showing that significant differences were not coupled with changes in the cut-off frequency within a session. . Compared to the predictable target, PSY was found to increase significantly more during the second day for the unpredictable target (target x day interaction). Compared to the predictable target, and the first assessment, no significant differences were found among trials (target x trial and day x trial interactions) for the performance descriptors. All factors interactions were also not significant.

Further analysis of reliability of the performance descriptors showed that for the predictable task, these measures were good (PSY=0.68, GX =0.72, GY =0.71) to excellently reliable (PSX =0.87). When using the unpredictable target, reliability was good (PSX=0.64, PSY=0.46, GX=0.49 and GY=0.66).

Discussion

This study explored learning effects within and between sessions and reliability of mediolateral balance performance descriptors when using visual tracking tasks based on centre of pressure feedback; for both a predictable and an unpredictable target. Linearity between ML CoP and the target displacement was assessed using Coh measures. This measure showed a moderate to high linearity for the unpredictable and predictable VTTs, respectively, which allowed characterizing balance control with PS and G. Comparisons between MELBA tasks showed greater PS and higher G when tracking a predictable target. Significant learning effects for PS (greater PS) and G (greater amplitude) between sessions (target x day interaction) and for PS within session (target x trial interaction) for G were observed when tracking an unpredictable target. Tracking accuracy and performance improvements were also significantly reflected in some of the performance descriptors (PSX and PSY), which, nevertheless, exhibited fair to good reliability. Lower reliability of the descriptors when using the unpredictable target can be explained by the significant learning effects observed in the PS and G measures.

Overall and for both targets (predictable and unpredictable), PS and G measures declined with increasing frequency content. This demonstrates that despite the simplicity of the task, it is challenging enough to observe a decline of the mediolateral balance performance in young healthy subjects, quantified by the cut-off frequencies (PSX and GX). This suggest the potential of MELBA as a balance assessment method in community-dwelling older adults and able-bodied population, since it is not likely to suffer from ceiling effects as observed with most of the currently available tools [5,242].

The first part of the MELBA test, with a predictable target, allowed assessment of physical tracking capacities, as the complexity of the task was minimized by using predictable traces and timing of the target signal and including an auditory cue (sensory redundancy). It is probable that, due to the predictable nature of this task, reliance upon feedforward control for guiding the task is increased whereas feedback control remains in place for sensing outcomes of the motor commands executed. As performance was similar over sessions for this specific task, and its descriptors (PSX, PSY, GX and GY) exhibited a fair to excellent reliability, this task seems a good measure of physical capacity: the capability to control ML balance without strong dependence on reactive control to sensory inputs, which would be more challenged by the unpredictable target.

When tracking the unpredictable target, reliance on feedback control of balance is expected to be predominant which may also account for lower performance in this task compared to the predictable target [21,207]. For instance, the continuous visual inputs processing induce a visuomotor delay, which may be responsible for a larger PS, compared to the predictable target. Gain decreases, on the other hand, may be the result of amplitude scaling in order to compensate for PS increases. Therefore comparatively lower PS and G demonstrate that despite physical resources to control balance are available, as shown when tracking the predictable target, visuomotor delay may constrain its utilization. This highlights the importance of cognition when producing motor commands to track an unpredictable target [18].

MELBA utilized visual tracking to assess balance by triggering voluntary balance responses when dealing with a constantly changing visual stimulus, especially for the unpredictable target. However, significant within (trial effect) and between sessions (day effect) learning effects using this target were found. Some available video games use CoP displacements as means to control the game. Such videogame-based interventions can engage elderly in balance training and improve the compliance towards their rehabilitation or prevention program [250]. Moreover, balance and fall risk assessment based on virtual reality environments or videogames have shown to be valid and to discriminate between fallers and non-fallers [250,251]. However, most of the measures used are related to functional scores and do not give insight in the underlying mechanisms. MELBA can provide a more objective measure of the balance performance by measuring phase-shift and gain in the frequency domain.

There are some limitations that need to be addressed. First, standing position and target displacements were normalized to body height and ankle width with pelvis and hip widths not considered in the model which may have affected performance [208,209,246]. For example, when dealing with ground ML perturbations, hip torque is lower at wide compared to narrow stance for a similar CoM displacement which requires adjustment of neural feedback gains [208]. This may affect performance when assessing older adults whom it is known to exhibit a wider stance [246]. Another limitation is that participants may have implemented different strategies to track the target. We chose to provide feedback of CoP given its ease of use. While subjects were explicitly instructed to maintain body alignment, which would impose a direct relationship between CoP and body center of mass (CoM) trajectories, failure to comply with this instruction may have affected task difficulty when CoM displacements were minimized for a given CoP displacement. Especially at higher frequencies, the variability of PS and G within and across participants increased. This might reflect such changes in movement strategy with increasing frequencies. Further study will explore this CoM/CoP relationship when tracking at a range of frequencies and its effect o performance as measured with MELBA. However, preliminary data not presented in the current paper show that in fact, CoM and CoP do relate in this tracking task. However, and as is to be expected, as frequency increases CoM displacement is reduced yet remains coherent with CoP movement. .

Current validation of commercially available force sensors (i.e. video game forceplate controllers) for clinical assessment may introduce opportunities for making MELBA a clinical tool [252]. The data processing as well as interpretation could be obtained from a simplified version of the software utilized in this experiment which can be easily installed in any computer. We are currently collecting data from older adults in order to determine reference parameters from healthy older adult population. Summarizing, MELBA has potential as a ML balance assessment method and training tool. The predictable target (and its descriptors) offers insight in the maximal physical capacity of the balance system to deal with the tracking task whereas the unpredictable target can provide information on the underlying cognitive mechanisms and sensory integration for controlling balance. In addition, learning effects found when using the unpredictable target indicate that balance performance can be improved. Further studies are needed to explore the use of MELBA to quantify the effect of ageing in ML balance performance, its sensitivity to sensory manipulations, underlying tracking mechanisms, discriminative capacity between people with and without balance problems, correlation with history of falls and clinical relevance as an assessment method and training tool.

Conclusions

Performance in tracking a predictable target with the CoP was higher compared to tracking an unpredictable target. This may indicate higher reliance upon cognitive mechanisms for the unpredictable target, which causes a smaller phase-shift and lower gain. Performance measures in tracking a predictable target may be useful to assess maximal physical capacities on the mediolateral balance task. Performance descriptors derived from the linear transfer function between target and CoP signals provide fair to excellently reliable descriptors of balance control. Learning effects observed when using the unpredictable target may indicate MELBA's potential as a balance training tool.

IV

Centre of pressure or centre of mass feedback in mediolateral balance assessment

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Abstract

The mediolateral balance assessment method (MELBA) consists of tracking a sinusoidal or multisine target with the centre of pressure as feedback (CoP_{fb}). The aim of the CoP trajectory is to elicit weight-shifting, i.e. movement of the centre of mass (CoM). However, it is not known whether CoP_{fb} elicits consistent mediolateral displacements of the CoM, whether CoM feedback (CoM_{fb}) is required to achieve this and whether CoP_{fb} or CoM_{fb} elicit different kinematic strategies. The aims of this study were to determine (1) the extent to which CoP imposes CoM displacements (CoM_d) during CoP_{fb}, (2) whether larger CoM_d are elicited by CoM_{fb} and (3) whether different kinematic strategies arise when using CoP_{fb} or $CoM_{fb.}$ Nineteen young adults performed MELBA with CoP_{fb} and CoM_{fb} from which coherence, gain and phase-shift between CoP-CoM and leg-trunk kinematics were calculated. CoM_d and CoP_d and leg and trunk excursions were also calculated. Results show that for CoP_{fb} tasks, CoP-CoM coherence was high, while the gain dropped with increasing frequency. The drop in gain was highly consistent between subjects. Reasonable trunk-leg coherence (≈0.6) was found over all frequencies and tasks. The leg-trunk angle gain increased with frequency in all tasks and was significantly higher in the CoM_{fb} compared to the CoP_{fb}. Significant interaction indicated that this difference increased with frequency. CoP_{fb} in MELBA elicits consistent CoM_d. However, different kinematics are employed in CoM_{fb} with more trunk movement and an ankle-to-hip shift as frequency increases. Hence CoM_{fb} may be preferable over CoP_{fb} despite the larger measurement effort involved.

Introduction

Impairments of balance in the mediolateral (ML) direction, reflected in inabilities to correctly shift weight and in impaired stepping responses are of special interest since these are associated to an increased number of falls [6,170]. Recently a mediolateral balance assessment method based on tracking of predictable and unpredictable target signals with the centre of pressure (CoP), coined MELBA, has been proposed [225] . MELBA characterizes balance control through the phase-shift (PS) and gain (G) between the CoP and a target signal that moves mediolaterally under a predictable (sinusoidal) or unpredictable (multisine) pattern. From these measures the frequency at which PS and G drop below a pre-defined threshold and the averages within the bandwidth defined by these frequencies are calculated. The method was shown to be reliable and did not show ceiling effects, not even not among young adults [225].

During locomotion as well as during transitions (i.e. rising from a chair) and standing, stability of the CoM has to be maintained through voluntary and reflexive motor commands to avoid falling [243]. The use of CoP_{fb} in balance testing therefore relies on the assumption that consistent ML CoM displacements (CoM_d) are elicited by ML CoP displacements (CoP_d), as the CoM is the controlled variable in balance control [253]. Since the distance between CoP and CoM ($\Delta_{CoP\text{-}CoM}$) is roughly proportional to the CoM acceleration (A_{CoM}), for limited angular excursions in upright stance a consistent relationship is expected albeit with CoM_d decreasing at constant CoP_d as frequency increases [254]. Although CoP_{fb} during MELBA tasks can thus impose consistent CoM_d, control over CoP may not arise as intuitively as control over the CoM, hence centre of mass feedback (CoM_{fb}) may be more suitable when demanding CoM_d. Furthermore, it is possible that CoP_{fb} and CoM_{fb} may elicit different strategies to control the CoM, which may be of utility in identifying the source of balance impairment at the effector levels. Therefore, a modified version of MELBA used CoM_{fb}. This was shown to be reliable and sensitive to age effects[243]. However, CoP feedback (CoP_{fb}) may be preferable in view of the instrumentation required.

The aim of this study therefore was to determine the extent to which CoP imposes CoM displacements (CoM_d) during CoP_{fb}, whether larger CoM_d are elicited by CoM_{fb} and whether different kinematic strategies arise when using CoP_{fb} or $CoM_{fb.}$ The results of this study will help to improve MELBA and its utility to determine ML balance impairments in older adults.

Methodology

Participants

Nineteen healthy young adults (11 women and 8 men, age: 26±3 years; height: 1.71±.09 m; weight: 67.2±12 kg) participated in this study. Participants were excluded if they presented any musculoskeletal or neurological condition. This study was approved by the local Ethical Committee in accordance with the ethical standards of the declaration of Helsinki. All participants were informed of the experimental procedures and signed informed consent before the experiment.

Task and Procedure

Each participant performed a series of ML CoP_{fb} and CoM_{fb} tracking tasks, while standing barefoot and with the arms crossed (for details of the set-up refer to Chapters III and V). CoP data were obtained using a Kistler-9281B force plate (Kistler Instruments AG, Winterthur, Switzerland) sampling at 60 samples/s. Body CoM was calculated with a 9-markers frontal plane model (forehead, shoulder, anterior-superior iliac spines, knees and ankles) tracked with an Optotrak Certus system (NDI, Waterloo, Canada). Gender specific CoM calculations were performed using scaling of anthropometric data and inertial parameters described by de Leva [215]. D-flow 3.10.0 software (Motek Medical, Amsterdam, The Netherlands) was used to produce target signals as well as to record (60 samples/s) and display target and either CoP or CoM data on a screen 2.5 m in front of the participant. ML-CoM tracking consisted of tracking a predictable and an unpredictable target signal using the ML displacement of the CoP or CoM projected on the screen. The target signal and CoP or CoM were represented by white and red spheres of 11 and 9 cm diameter, respectively.

The *predictable* target signal was constructed using 2 blocks of 20 seconds, 1 block of 10 seconds and 17 blocks of 5 seconds, each composed by one sine wave, which increased in frequency from 0.1 to 2.0 Hz in steps of 0.1 Hz. The total duration for this target signal was 135 seconds. The *unpredictable* target signal was constructed using 15 blocks composed by the sum of 6 consecutive sine waves separated by 0.1 Hz. A pseudorandom phase-shift between sine waves between -1 to 1 period was introduced in order to avoid predictability. After each block the lowest frequency, which started at 0.1 Hz, was increased by 0.1 Hz until it reached 1.5 Hz. Duration was 40s for block 1, 20s for block 2, 10s for block 3, 8s for blocks 4 and 5, 6s for blocks 6 and 7, and 4 seconds for blocks 8 to 15. The total duration for this target signal was 132 seconds.

Each participant performed 6 CoP and 6 CoM tracking trials: 3 with the predictable and 3 with the unpredictable target for each type of feedback provided (CoP_{fb} and CoM_{fb}). One practice trial was allowed for each of the conditions. Target maximum side-to-side displacement for both, predictable and unpredictable targets, was normalized for each subject at 100% of the between-heels distance when using CoP_{fb} and 50% when using CoM_{fb}. These distances were chosen based on pilot experiments, which showed that subjects were unable to move CoM as far as CoP in the ML direction during MELBA tasks using CoM_{fb}. On average, the participants stood on the force plate with 19.0 SD 1.0 cm distance between heels, which determined a maximum target displacement of 18.9 SD 1.1 cm when using CoP_{fb} 9.5 SD 0.5 cm when using CoM_{fb}.

Data analysis

CoP-CoM relationship

CoP-CoM relationship over the frequency ranges in the target signal was described by the gain of the linear constant coefficient transfer function between CoP_d and CoM_d from which gain (G) and coherence (Coh) were calculated. G values < 1 for the CoP-CoM relationship will indicate a lower magnitude of the ML CoM_d in response to ML CoP_d. Coh was used to determine linearity between CoP and CoM. Perfect linearity yields Coh = 1 over all frequencies comprising the target signal. CoP_d and CoM_d were calculated over the time windows described above. These measures were used to compare the amount of CoP_d and CoM_d imposed when having CoP_{fb} and CoM_{fb} in both, predictable and unpredictable tasks.

Movement strategies

Legs (Φ_{leg}) and trunk (Φ_{trunk}) angles relationship over the frequency ranges in the target signal was described by the gain of the linear constant coefficient transfer function between Φ_{legs} and Φ_{trunk} from which gain (G) and coherence (Coh) were calculated. G < 1 for the legs- trunk angles relationship will a lower magnitude of the Φ_{trunk} in response to Φ_{legs}. Coh was used to determine linearity between Φ_{legs} and Φ_{trunk}. Legs ($legs_{ad}$) and trunk ($trunk_{ad}$) angular displacements were calculated over the whole trials and within the time windows described for the MELBA tasks. These measures were used to compare the amount of $legs_{ad}$ and $trunk_{ad}$ imposed when having CoP_{fb} and CoM_{fb} in both, predictable and unpredictable tasks.

Statistical Analysis

A multivariate ANOVA was performed to determine differences in PS, G and Coh between CoP-CoM and legs-trunk angles for the predictable and unpredictable MELBA tasks at each frequency

(0.1 to 2.0 Hz at steps of 0.1) between CoP_{fb} and CoM_{fb} (feedback) as well as the interaction between frequency and feedback. A multivariate ANOVA was also performed to determine differences in CoP_d, CoM_d, $legs_{ad}$ and $trunk_{ad}$ between targets and feedbacks (CoP_{fb} and CoM_{fb}). For all analyses significance level was set at p <.05. Statistical analyses were performed using IBM SPSS (Statistics 21).

Results

Averaged plots of CoP_d, CoM_d, $legs_{ad}$ and $trunk_{ad}$ during the CoP_{fb} and CoM_{fb} for both, predictable and unpredictable targets are presented in figure 1. Figures 2 and 3 present the CoP-CoM and legs-trunk G and Coh, respectively. Table 1 presents the results for the statistical tests performed to determine the effect of feedback (CoP_{fb} and CoM_{fb}) and frequency (0.1 to 2.0 Hz) on G and Coh. Table 2 presents means and results for the statistical tests performed to determine differences in CoP_d, CoM_d, $legs_{ad}$ and $trunk_{ad}$ between CoP_{fb} and CoM_{fb}.

Overall, using both feedbacks CoP-CoM coherence values show high linearity (>.7), however, Coh was significantly higher when using CoM_{fb}. Significant effects of frequency and feedback*frequency interaction on Coh were found. These differences were greater in the unpredictable target. CoP-CoM G dropped with increasing frequency and was highly consistent between subjects. However, for the unpredictable target when using CoM_{fb} there was a significantly lower CoP-CoM G in the 0.1-0.8 Hz range. Significant effects of feedback*frequency interaction were also found for G. For the latter, a stepper drop was observed when using CoM_{fb} in the predictable target.

In relation to legs-trunk Coh, no effect of feedback was found for the predictable target, however, for the unpredictable this was significantly lower when using CoP_{fb}. A significant effect of frequency was found for G and Coh when tracking both targets. Whereas an interaction effect was found for G and Coh in the unpredictable task, this interaction was only present for G in the predictable task.

Significantly greater CoP_d, CoM_d, and $trunk_{ad}$ were found when using CoM_{fb} and when tracking the predictable target. Significantly greater $legs_{ad}$ target in CoM_{fb} was only found when tracking the predictable. A significant target*feedback effect was found for CoM_d, $legs_{ad}$ and $trunk_{ad}$ but not for CoP_d.

Discussion

This study primarily aimed to determine the extent to which CoP_{fb} imposes consistent CoM displacements (CoM_d) . CoP-CoM coherence values show a high linearity in the response of CoM to CoP displacements. This response, however, is scaled with frequency content of the target signals, with higher frequencies imposing larger CoM acceleration [254], as is reflected in the consistent drop in CoP-CoM gain.

The second aim of this study was to determine whether larger CoM_d are elicited by CoM_{fb} when compared to CoP_{fb}. CoP_{fb} elicited smaller CoM_d and CoP_d than CoM_{fb} even when the side-to-side maximum CoP_d demanded in the CoP_{fb} tasks was double than in the CoM_{fb}. This shows that to challenge the balance control system by increasing the demands of CoM_d, direct CoM_{fb} is preferable. Furthermore, a greater G and Coh when using CoM_{fb} may indicate that subjects were more responsive to the demands of the tracking tasks than when using CoP_{fb}. It is noteworthy, however, that in CoM_{fb} G is lower at the lowest frequencies, especially in the unpredictable tracking task. CoP_{fb} may function relatively better than CoM_{fb} at low frequencies, because the relation between CoP_d and CoM_d is closer to proportional. CoM_{fb} demands larger CoP_d to displace CoM than CoP_{fb}. This involves similar contribution of $legs_{ad}$ and $trunk_{ad}$ during CoM_{fb} in contrast to CoP_{fb} where $legs_{ad}$ strategy is predominant. Larger CoP_d during CoM_{fb} result in a lower CoP-CoM gain when compared to CoP_{fb} at the 0.1-0.3Hz range.

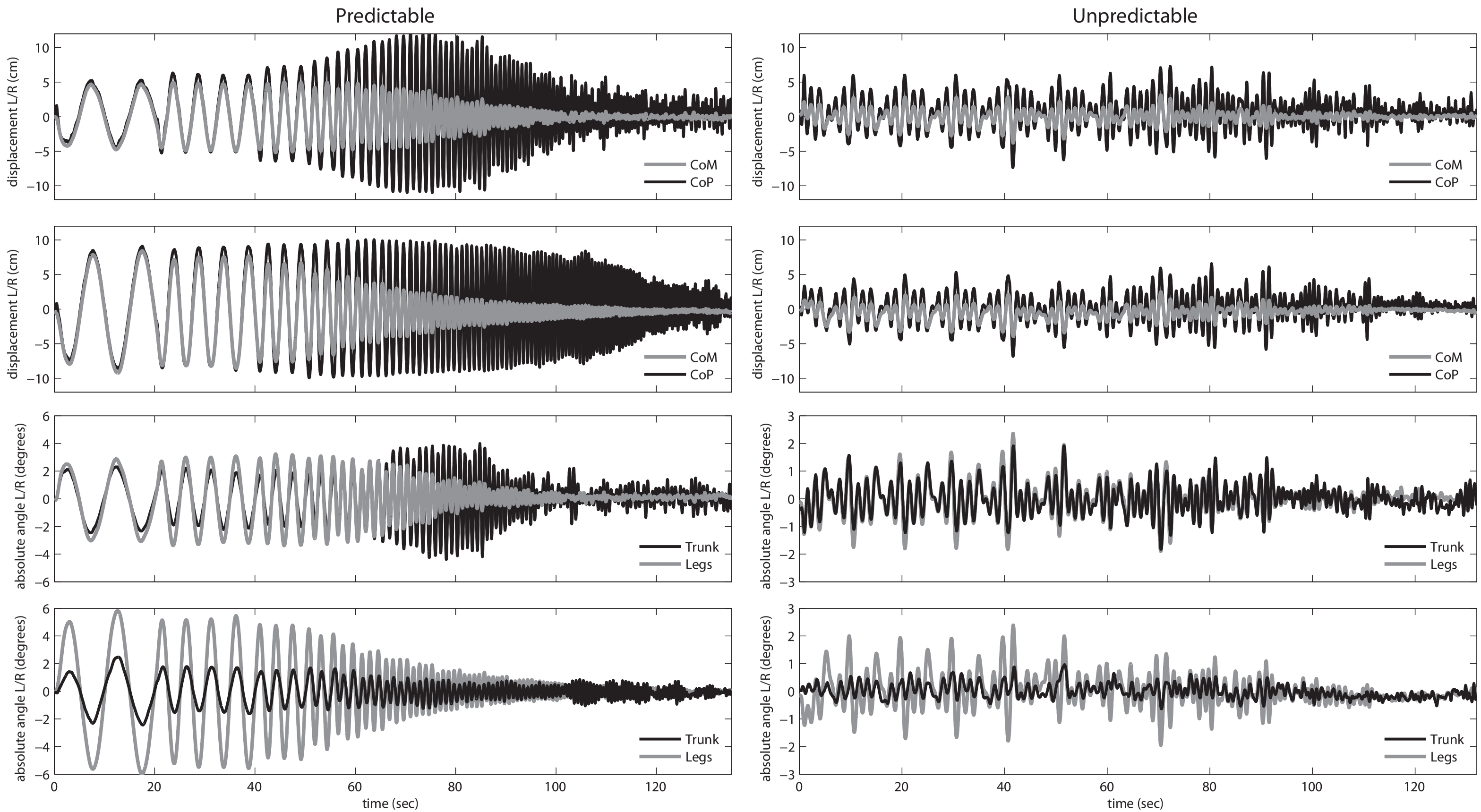

Figure 4.1. Averaged plots for CoM and CoP displacements (meters) during both MELBA tasks (predictable on the left panel and unpredictable on the right panel) when using CoM_{fb} (first row) and CoP_{fb} (second row). Averaged plots for leg and trunk angles (degrees) during both MELBA tasks when using CoMfb (third row) and CoPfb (fourth row) are also presented.

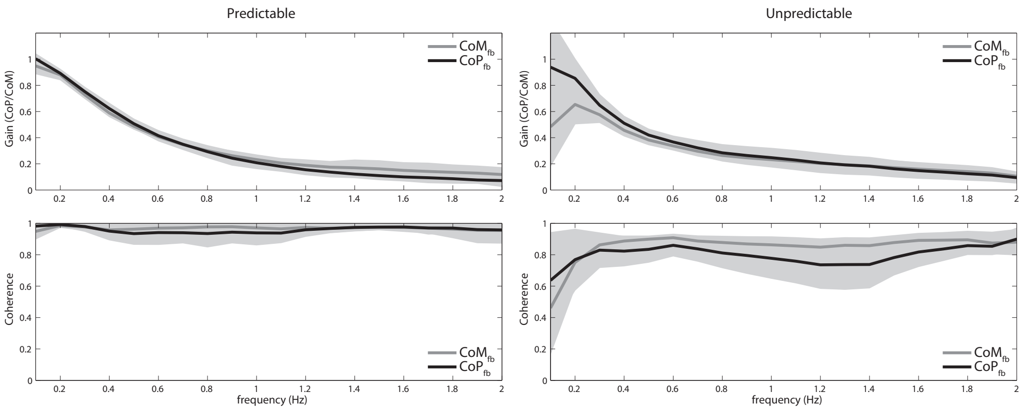

Figure 4.2. Averaged plots for CoP-CoM PS, G and Coh for the predictable (left panel) and unpredictable (right panel) tracking tasks when using CoPfb (black line) and CoMfb (dark grey line). Shaded light grey area represent ±sd.

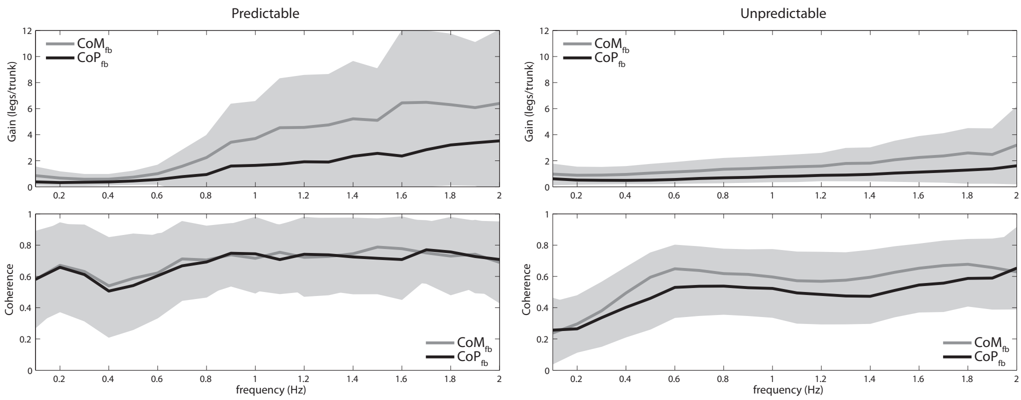

Figure 4.3. Averaged plots for leg-trunk PS, G and Coh for the predictable (left panel) and unpredictable (right panel) tracking tasks when using CoPfb (black line) and CoMfb (dark grey line). Shaded light grey area represent ±sd.

Table 4.1. Results for the multivariate ANOVAs tests performed to determine the effect of feedback (CoPfb and CoMfb) and frequency (0.1 to 2.0 Hz at steps of 0.1) as well as interaction effect on CoP-CoM and leg-trunk PS, G and Coh for both, predictable and unpredictable target. Significant differences are presented in bold.

		Predictable		**Unpredictable**	
		G	Coh	G	Coh
CoP-CoM	feedback	**<.01**	**<.01**	**<.01**	**<.01**
	frequency	**<.01**	**<.01**	**<.01**	**<.01**
	feedback*frequency	**<.01**	**<.01**	**<.01**	**<.01**
leg-trunk	feedback	**<.01**	.18	**<.01**	**<.01**
	frequency	**<.01**	**<.01**	**<.01**	**<.01**
	feedback*frequency	**<.01**	.98	**<.01**	**.05**

Table 4.2. Descriptive statistics (mean±sd) and results for the statistical tests performed to determine differences in CoP_d, CoM_d, leg_d and $trunk_d$ between targets (predictable and unpredictable) and feedback (CoP_{fb} and CoM_{fb}) as well as interaction effect. Significant differences are highlighted in bold.

	Predictable				**Unpredictable**						
	CoM_{fb}		CoP_{fb}		CoM_{fb}		CoP_{fb}				
	mean	sd	mean	sd	mean	sd	mean	sd	tar	fb	tar*fb
CoM_d	10.8	1.5	9.3	1.1	7.2	1.0	4.9	1.0	**<.001**	**<.001**	**.020**
CoP_d	45.8	7.3	40.4	4.3	38.0	7.9	29.9	4.7	**<.001**	**<.001**	.101
leg_{ad}	554.0	159.5	597.0	113.7	337.3	65.5	291.3	75.4	**<.001**	.916	**.003**
$trunk_{ad}$	1592.3	944.4	769.2	759.2	579.5	282.7	290.9	167.3	**<.001**	**<.001**	**.001**

The third aim of this study was to determine whether different kinematic strategies arise when utilizing CoP_{fb} and CoM_{fb}. Significantly larger $legs_{ad}$ and $trunk_{ad}$ and the frequency dependent ratio between the two when using CoM_{fb} show that a wider variety of motor strategies are called into play than when using CoP_{fb}. This may also indicate a greater challenge for the balance control system, since kinematic strategies shift from ankle to hip-trunk muscles as demands of the tracking tasks increase. Since an age-related proximal-distal shift in locus of function has previously been shown in gait [255], an earlier strategy shift (increased hip muscle activity) in MELBA using CoM_{fb} may indicate deterioration of distal neuromuscular function in the older adults, such as reduced muscle strength at the ankle joint.

The use of CoP_{fb} in the context of geriatric assessment or clinical settings may be preferable over CoM_{fb} given the lower costs and lesser requirements with respect to time and equipment [145]. However, to make sure that the test is sufficiently challenging for older adults, who may exhibit only minor impairments of balance, the greater demands in terms of CoM_d, and $trunk_{ad}$ that are imposed using CoM_{fb} may better reflect maximal capabilities of the balance control system than CoP_{fb} tracking tasks [256]. Although CoM_{fb} may be cumbersome to be implemented at present, current developments of markerless motion capture systems are likely to allow simpler implementation in the near future [257].

Conclusions

CoP_{fb} in MELBA elicits consistent CoM_d. However, different kinematics are employed in CoM_{fb} with more trunk movement and an ankle-to-hip shift as frequency increases. Hence CoM_{fb} may be preferable over CoP_{fb} despite the larger measurement effort currently involved.

V

Age effects on mediolateral balance control

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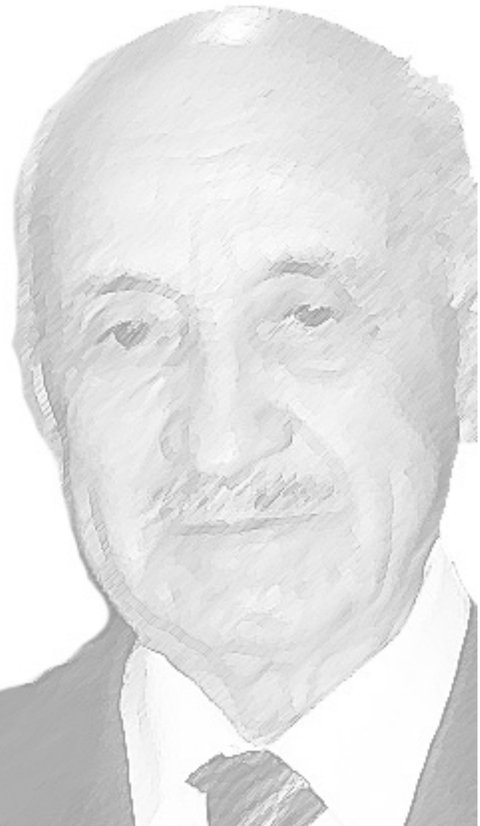

Abstract

Age-related balance impairments, particularly in mediolateral direction (ML) may cause falls. Sufficiently sensitive and reliable ML balance tests are, however, lacking. This study is aimed to determine (1) the effect of age on and (2) the reliability of ML balance performance using Center of Mass (CoM) tracking. Balance performance of 19 young (26±3 years) and 19 older (72±5 years) adults on ML-CoM tracking tasks was compared. Subjects tracked predictable and unpredictable target displacements at increasing frequencies with their CoM by shifting their weight sideward. Phase-shift (response delay) and gain (amplitude difference) between the CoM and target in the frequency domain were used to quantify performance. Thirteen older and all young adults were reassessed to determine reliability of balance performance measures. In addition, all older adults performed a series of clinical balance tests and conventional posturography was done in a sub-sample. Phase-shift and gain dropped below pre-determined thresholds (-90 degrees and 0.5) at lower frequencies in the older adults and were even lower below these frequencies than in young adults. Performance measures showed good to excellent reliability in both groups. All clinical scores were close to the maximum and no age effect was found using posturography. ML balance performance measures exhibited small but systematic between-session differences indicative of learning. The ability to accurately perform ML-CoM tracking deteriorates with age. ML-CoM tracking tasks form a reliable tool to assess ML balance in young and older adults and are more sensitive to age-related impairment than posturography and clinical tests.

Introduction

It is widely accepted that, in our aging society, falls and fall-related injuries are a major problem with high personal and economic impact [7]. Balance impairments form one of the main risk factors for falls, not only in patient populations but also in community-dwelling older adults [163]. Most of the individuals older than 60 years exhibit some degree of balance impairment, which gradually affects mobility and increases dependency [258]. Therefore, early and adequate assessment of balance impairments is of paramount importance to identify those individuals in need of preventive care [259] and to monitor effects of preventive interventions [128].

Mediolateral (ML) balance impairments have in particular been associated with an increased risk of falling in the older population [151,156,169]. For instance, in prospective and retrospective studies, postural sway parameters in the ML direction have been shown to be higher (i.e. larger area and excursion of the centre of pressure) in fallers than in non-fallers [151]. Nevertheless, as balance control declines gradually with aging, current clinical tools are not sensitive enough to detect early stage impairments in community-dwelling older adults, as these tests exhibit ceiling effects [128]. For instance, Berg and POMA scales have shown ceiling effects even in older adults who exhibit moderate to severe limitations of function (i.e. inability to climb stairs without assistance)[242]. Also conventional posturography, does not consistently discriminate between young and older adults [158]. It appears that ability of balance performance measurements to predict fall risk can be improved over that of conventional posturography by adding a more dynamic component, which involves center of mass (CoM) movements or weight shifting [260]. In line with this, slow lateral stepping responses have been associated with fall risk in older adults [156] and based on videos of real-life falls, inadequate weight shifting accounted for 41% of the falls [6]. Although the latter study focused on older adults living in long-term care facilities, previous studies in community-dwelling older adults also suggest that a considerable proportion of falls can be attributed to incorrect weight-shifting or daily-life tasks that challenge ML balance [173,261].

Sufficient sensitivity to detect age-related impairments in ML balance control, even in relatively fit and healthy community-dwelling older, can be reached by utilizing tests with incremental difficulty, which can probe the limits of the responsiveness of the balance control system in relation to the demands of the task. The responsiveness can be expressed as control bandwidth, i.e. the range of frequencies over which one can operate within some tolerated error level. For example, a low frequency sinusoidal target signal can be tracked closely, but as the frequency of the signal increases, limits in control bandwidth result in growing tracking errors. Bandwidth of ML balance control can be reduced by slower central and peripheral processing of sensory information [92] and reduced ability to execute motor commands due to muscle weakness (reduced strength and power) [74].

Recent work by our group showed that a mediolateral balance assessment task (coined MELBA), using the center of pressure (CoP) for tracking a visual target allows determining limits in control bandwidth even in healthy young adults [225]. In the current study, we used a modified version of MELBA, in which the subject tracks a target with his or her body CoM, instead of CoP. We believe that using CoM instead of CoP is more meaningful and intuitive, since the CoM is the controlled variable in balancing and weight shifting [262].

The aim of this study was to determine the effect of age on balance responsiveness (control bandwidth) using MELBA. We hypothesized that older adults would have a narrower control bandwidth than young adults. To compare sensitivity of MELBA with conventional methods, we also used posturography. In addition, we investigated test-retest reliability of the modified MELBA. Based on results obtained with CoP tracking [225], we hypothesized that test-retest reliability would be similar or better than CoP-tracking.

Methods

Participants

Nineteen healthy older and 19 healthy younger subjects were recruited for this study. To further characterize the older participants, the mini mental state examination MMSE, the Quickscreen (QS)[263], short physical performance battery (SPPB)[200], Berg balance scale (BBS)[264], miniBEST test (MB)[265], performance-oriented mobility assessment balance section (POMA-B) [266] and timed up-and-go (TUG)[267] were used. Performance during the timed up-and-go with dual task (DTUG) was extracted from the MB. This research was approved by the Ethical Committee of the Faculty of Human Movement Sciences, VU University, Amsterdam (2011-48M), in accordance with the ethical standards of the declaration of Helsinki. All participants were informed of the experimental procedures and signed informed consent was obtained prior to the experiment.

Task and Procedure

Each participant performed a series of ML-CoM tracking tasks, while standing barefoot and with the arms crossed in a quiet and low-intensity lit room (for set-up details, see Figure 5.1). Body CoM was calculated with a 9-markers frontal plane model (forehead, shoulder, anterior-superior iliac spines, knees and ankles) using an Optotrak Certus motion capture system (Northern Digital Instruments, Canada). Gender specific CoM calculations were performed using scaling of anthropometric data and inertial parameters described by de Leva [215]. D-flow 3.10.0 software (Motek Medical, The Netherlands) was used to produce target signals as well as to record (60 samples/s) and display target and CoM data on a screen 2.5 m in front of the participant. ML-CoM tracking consisted of tracking a predictable and unpredictable target signal using the ML displacement of the CoM projected on the screen. The target signal and CoM were represented by white and red spheres of 11 and 9 cm diameter, respectively. CoP data were collected using a Kistler-9281B force plate (Kistler Instruments AG, Winterthur, Switzerland) sampled at 60 samples/s.

The *predictable* target signal was constructed using 2 blocks of 20 seconds, 1 block of 10 seconds and 17 blocks of 5 seconds, each composed by one sine wave, which increased in frequency from 0.1 to 2.0 Hz in steps of 0.1 Hz. This information was enhanced using a metronome synchronized with the maximum displacement of the target in order to increase sensory input abundance. The total ML-CoM tracking time for this target signal was 135 seconds.

The *unpredictable* target signal was constructed using 15 blocks composed by the sum of 6 consecutive sine waves separated by 0.1 Hz. A pseudorandom phase-shift between sine waves between -1 to 1 period was introduced in order to avoid predictability. After each block the lowest frequency, which started at 0.1 Hz, was increased by 0.1 Hz until it reached 1.5 Hz. Duration was 40s for block 1, 20s for block 2, 10s for block 3, 8s for blocks 4 and 5, 6s for blocks 6 and 7, and 4 seconds for blocks 8 to 15. Duration of the blocks was chosen to obtain a minimum of 2 cycles per frequency contained in the block. The total ML-CoM tracking time for this target signal was 132 seconds. Examples of the two target signals are depicted in Figure 5.1.

Each participant performed 6 ML-CoM tracking trials: 3 with the predictable and 3 with the unpredictable target. Before performing the test, one practice trial was allowed for each of the conditions. To determine test-retest reliability, all younger and 13 of the older adults repeated the test in a second session 7 days later at the same time of the day. Trials were performed with at least 1 minute of rest in between. Since stance width alters lower limb neuromechanical responses when displacing CoM and CoP in the ML direction [208], stance width was standardized by setting the heel distance to 11% of body height. A fixed 14° stance angle was

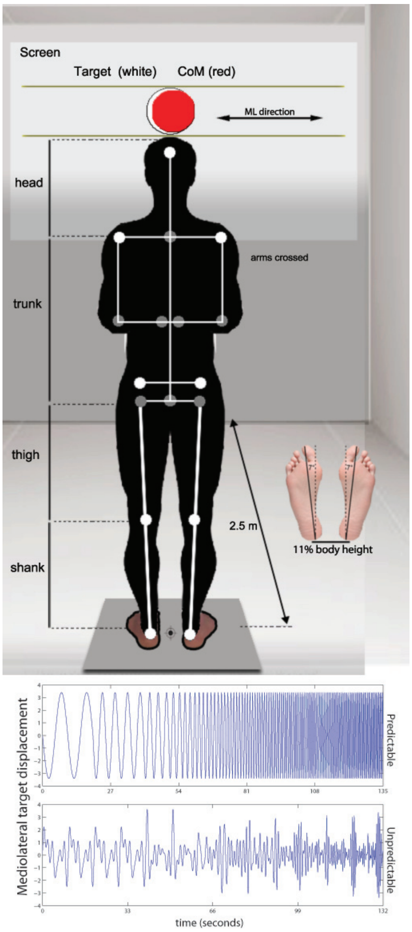

Figure 5.1. Illustration of the set-up and the model for Centre of Mass (CoM) calculation utilized in this experiment, showing a silhouette of a subject standing in the middle of a forceplate with marker placement superimposed (in white actual makers and in grey estimated joint centres) and the display of the CoM feedback (red sphere). The white sphere in the centre represents target which moved in the mediolateral (ML) direction following the patterns depicted in the bottom panel: predictable (top) and unpredictable (bottom). An insertion of foot soles is presented showing foot positioning during the experiments (stance width and angle).

used across all participants (Figure 5.1). These stance measures have been shown to be within the values of normal stance [246]. Target maximum side-to-side displacement for both target signals was normalized for each subject at 50% of stance width; allowing ML-CoM displacements to be within the base of support. On average, older participants stood on the force plate with 19.0x±1.0 cm distance between heels, which determined a maximum target displacement of 9.5x±0.5 cm whereas younger participants stood on the force plate with 18.9.x±1.1 cm distance between heels, which determined a maximum target displacement of 9.4±0.5 cm. Between groups displacement differences were not significant. Additionally, a subsample of 10 older adults and all younger participants performed 3 standing still trials of 50 seconds with the eyes open and 3 with eyes closed for comparison with ML-CoM non-tracking postural sway measures and conventional posturographic measures (i.e. CoP sway area). No data was discarded and the use of subsamples for the re-test session and posturography measures was imposed by the time constraints of the participants who were unable to attend two sessions.

Data analysis

All data analysis was performed using custom-made software in Matlab R2011a (Mathworks, Natick MA, USA). Balance performance over the frequency ranges in the target signal was described by the gain of the linear constant coefficient transfer function between CoM and target signal. This analysis was performed using the Welch algorithm over windows of 0.25 times the length of the target (per block) with 90% overlap between windows [225]. For the unpredictable target, phase-shift, gain and coherence were calculated as the average of the values at each frequency over blocks with overlapping frequency content. The phase-shift (PS) reflects the delay (in degrees) between target and CoM whereas gain (G) reflects the ratio between the target and CoM amplitudes; both in the frequency domain. Perfect performance implies PS = 0 and G = 1 over all frequencies comprising the target signal. In addition, the coherence (Coh) was determined, as a measure of the correspondence between the target and CoM in the frequency domain, which in this study was used to corroborate the assumption of input (target)/output (CoM) linearity and therewith the validity of estimates of PS and G. Perfect linearity produces Coh = 1 over all frequencies comprising the target signal.

To characterize balance performance, 4 descriptors were calculated. First, the values at which PS dropped below 90 degrees and G dropped below 0.5 were determined as the cutoff frequencies (coined f_{PS} and f_G, respectively). Second, PS_{mean} and G_{mean} were computed as the average of the G and PS values within the bandwidth determined by f_{PS} and f_G, respectively.

For the posturographic measures (eyes open and eyes closed), CoP sway area and mean velocity, maximal velocity, total excursion and standard deviation of the CoP in the anterioposterior (AP) and ML directions were calculated. Additionally the sum of energies across the .05-2.0 Hz power spectrum of the ML-CoM postural sway was analyzed. This range was chosen since it contains the frequencies present in both targets used in the tracking tasks. Although conventional posturography uses CoP to asses balance, it has been shown that during unperturbed upright standing there is a direct relation between CoP and CoM [262].

Statistical Analysis

Repeated measures ANOVAs were performed on the dependent variables f_{PS}, PS_{mean}, f_G, and G_{mean} with age as a between-subject factor (younger versus older), and target (predictable and unpredictable target) as a within-subject factor. For this analysis the averaged values over three trials performed in session 1 were used. The strength of the age-effect was quantified by calculating the effect size (eta squared).

To analyze test-retest reliability, the data of all subjects participating in both sessions were used. First, to assess systematic differences, a repeated measures ANOVA with age as a between-subject factor (younger versus older), target (predictable and unpredictable target), trial number (1 to 3) and session (1 or 2) as within-subject factors. In view of multiple testing, α was set at .0125 (.05/4). To determine reliability of performance descriptors, intraclass correlations (ICC 2,1) of the measured variables were calculated for the whole group. To better determine reliability of the measures when applied in a specific age range, ICC was also performed for each age group separately. Measures were considered to exhibit excellent reliability when ICC > .74, good = .60-.74 and fair= .40-.59 [268].

A univariate ANOVA with age as a random factor was performed to determine the effect of age on ML-CoM non-tracking postural sway (conventional posturography). Separate univariate ANOVAs with age as a random factor were used to determine the effect of age on CoP sway measures with eyes open and closed. To better compare age effect on MELBA and conventional posturography, α was also set at 0.0125 and the effect size of age was quantified using eta-squared. Statistical analyses were performed using IBM SPSS (Statistics 21).

Results

Subjects

Demographics for all subjects and results of clinical balance tests for the older adults are presented in Table 5.1. No differences in height and weight were found between groups. Participants did not report any musculoskeletal or neurological condition or use of medication that could affect balance. The older adults scored close to the maximum in all clinical tests and scores were above the cut-off scores for the highest (best balance performance) category defined for each test.

Table 5.1. Top part of the table shows demographics for all participants. Bottom part of the table shows the descriptive statistics (mean, ±sd, median, lowest and highest scores) for the clinical measures of balance in the older participants: Quickscreen (QS), short physical performance battery (SPPB), Berg balance scale (BBS), miniBEST test (MB) and performance-oriented mobility assessment balance section (POMA-B). For the timed up-and-go (TUG) and dual-task timed up-and-go (DTUG), the mean ±sd and 95% confidence interval are presented.

				Older adults		Young	
				mean	sd	mean	sd
Demo-graphics		Age	(years)	72.0	4.6	26.0	3.3
		Height	(m)	1.7	.1	1.7	.1
		Weight	(kg)	76.6	15.2	67.0	12.0
				mean	sd	95% confidence interval	
Clinical measures	time	TUG	(seconds)	6.16	1.05	5.65	6.67
		DTUG	(seconds)	7.29	1.75	6.45	8.13
				median			
	scores	QS	(min 0)	2		0	4
		BBS	(max 56)	56		53	56
		SPPB	(max 12)	12		10	12
		MiniBEST	(max 28)	26		23	28
		POMA-B	(max 16)	16			

ML-CoM tracking

For all balance performance measures (f_{PS}, PS_{mean}, f_G and G_{mean}), significant main effects of age were found ($p<.001$), indicating a narrower control bandwidth in the older compared to the younger adults (Figure 5.2; Table 5.1). In addition, a significant main effect of target was found, with all measures exhibiting lower values when tracking the unpredictable target (Figure 5.2; Table 5.2). No interactions between age and target were found. Although lower than for the target main effect, the effect size of age for all measures was medium ($\eta^2 \leq 0.13$) to large ($\eta^2 \leq 0.38$).

A moderate to high linearity between ML-CoM and the displacement of both targets was found as expressed by mean coherences (0.1 to 2.0 Hz range) > 0.4 and > 0.6 for unpredictable and predictable ML-CoM tracking, respectively. This supports characterization of balance control using gain and phase-shift. Overall, subjects performed better when tracking the predictable target, reflected by gain values closer to 1 and phase shifts closer to 0, compared to tracking the unpredictable target, especially for input frequencies below 0.8 Hz. For the unpredictable target, near-optimal values for gain and phase were not observed, underlining the challenging nature of this task.

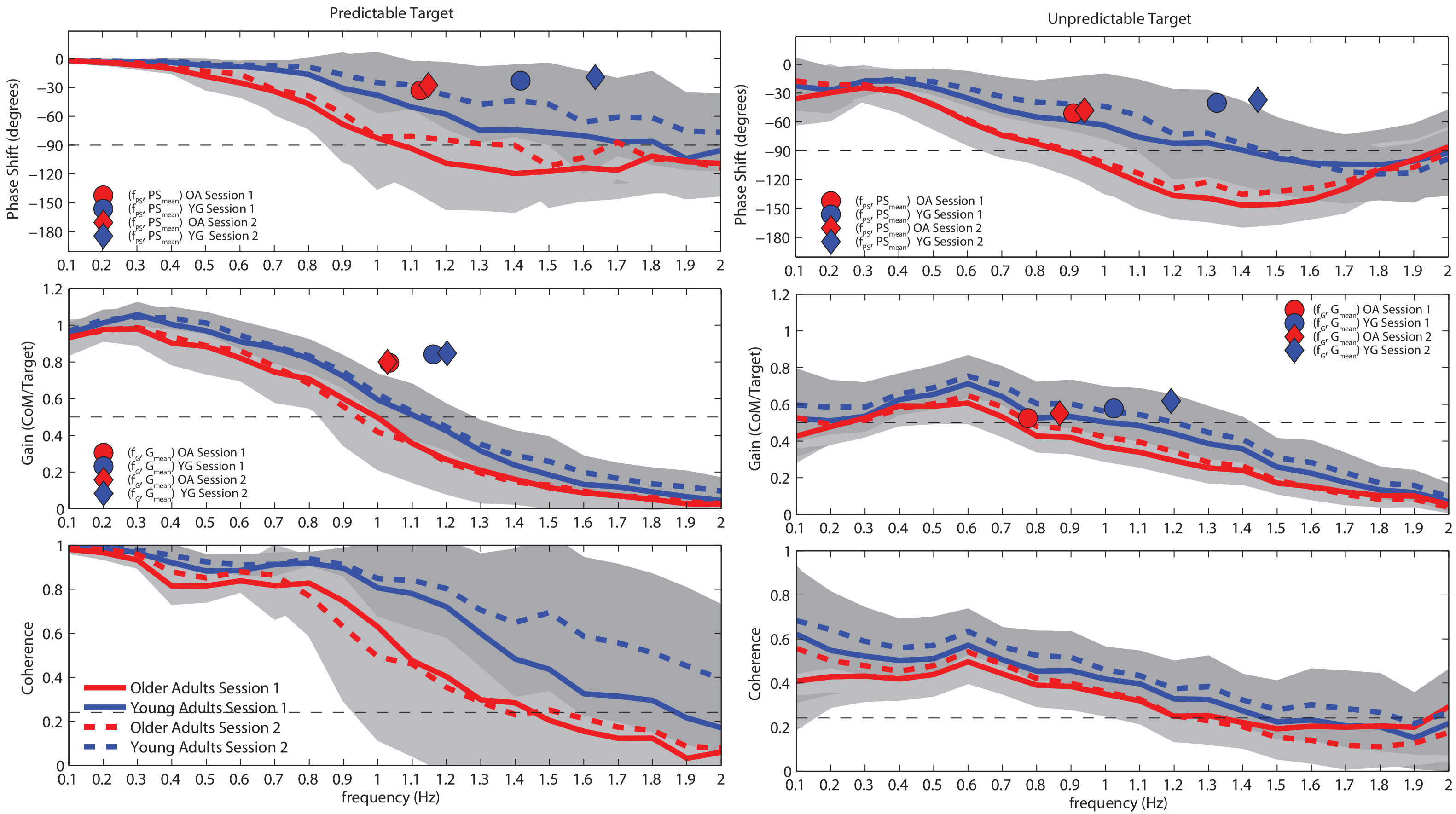

Figure 5.2. Averaged curves (± sd) for phase shift (top panel), gain (mid panel) and coherence (bottom panel) measures using both, predictable target (left) and unpredictable (right) targets, during first (continuous line) and second (dashed line) sessions and for the younger (in black) and the older adults (in dark grey). Grey shading indicates the ± sd for all subjects and for all trials. Markers inserted in the plots indicate means for performance descriptors for the first session (circular markers) and second session (diamond markers) for the younger (in black) and the older adults (in dark grey).

Table 5.2. Descriptive statistics for MELBA performance descriptors (f_{PS}, PS_{mean}, f_G and G_{mean}) for the predictable and unpredictable targets. Right part of the table summarizes the *p*-values and effect sizes (η^2) of the repeated measures ANOVAs for the between-subjects comparison ('age': older vs young adults) the main effect of target ('tar': unpredictable and predictable) and age-by-target interaction. Statistically significant *p*-values are presented in bold

		Predictable				Unpredictable				RMANOVA (effects of age and target in 1st session)					
		Session 1		Session 2		Session 1		Session 2		*p*			η^2		
		mean	sd	mean	sd	mean	sd	mean	sd	age	tar	tar* age	age	tar	tar* age
f_{PS} (Hz)	**Young**	1.42	0.34	1.64	0.32	1.33	0.39	1.45	0.34	**<.01**	**<.01**	.11	.26	.35	.08
	Older	1.13	0.31	1.15	0.33	0.91	0.17	0.94	0.2						
PS_{mean} (°)	**Young**	-23.04	7.08	-19.33	6.77	-40.18	10.07	-37.18	9.21	**<.01**	**<.01**	.72	.36	.87	<.01
	Older	-33	5.58	-27.58	5.38	-51.03	6.98	-47.89	9.53						
f_G (Hz)	**Young**	1.16	0.1	1.2	0.13	1.02	0.25	1.19	0.27	**<.01**	**<.01**	.14	.34	.45	.07
	Older	1.03	0.15	1.03	0.16	0.77	0.15	0.87	0.14						
G_{mean}	**Young**	0.84	0.05	0.85	0.04	0.58	0.08	0.62	0.08	**.01**	**<.01**	.83	.20	.92	<.01
	Older	0.79	0.06	0.8	0.04	0.52	0.06	0.55	0.07						

When testing over repeated sessions, significant main effects of session were found for all balance performance measures (all p≤0.01), with a slightly better performance during the second session (Figure 5.2). Furthermore, we found interactions of session and target for f_G and G_{mean} (p≤0.01), indicating more improved performance over sessions, when tracking the predictable target. A significant main effect of trial was found only for f_{PS} (p <0.01) with a consistent improvement over trials in the younger adults mainly, as indicated by an age-by-trial interaction (p =0.01). A significant interaction of trial and age was also found for PS_{mean} (p =0.01), here with the older adults exhibiting more improved performance over trials. Finally, a significant target-by-trial interaction was found for f_{PS} (p =0.01), with more improved performance over trials when tracking the unpredictable target. In spite of these systematic between-session effects, ICCs showed that for all subjects pooled, reliability of all balance performance descriptors was

Table 5.3. Intraclass correlations (absolute agreement) for the performance descriptors (f_{PS}, PS_{mean}, f_G and G_{mean}) for both visual tracking tasks (predictable and unpredictable) for all subjects and stratified by age group. Descriptors exhibiting excellent reliability are shown in bold.

	Intraclass correlations							
	Predictable				**Unpredictable**			
	f_{PS}	PS_{mean}	f_G	G_{mean}	f_{PS}	PS_{mean}	f_G	G_{mean}
All (32)	**.86**	**.83**	**.86**	**.77**	**.91**	**.88**	**.84**	**.87**
Young (19)	.74	**.83**	.74	.62	**.85**	**.89**	**.78**	**.85**
Older (13)	**.95**	.57	**.87**	.64	**.85**	.68	.74	**.85**

excellent, with ICC values ranging from 0.77 for G_{mean} when tracking the predictable target to 0.91 for f_{PS} when tracking the unpredictable target (Table 5.3). As expected, stratified analysis by age group showed lower ICC values, but reliability still ranged from fair to excellent.

Posturography

No age effect on ML-CoM non-tracking postural sway, as expressed by the energy across the 0.05-2.0 Hz range in quiet standing, was found (younger: 0.27 ±.09 m^2/Hz and older: 0.27 ±.22 m^2/Hz, *p*=0.91). In addition, no significant differences were found conventional posturography (CoP sway measures) measures. The largest effect sizes were found for the maximum sway velocity in the ML direction for both, eyes-open and eyes-closed conditions (with p = 0.03, NS after Bonferroni correction), with, however, lower velocities for the older adults.

Discussion

We studied the effects of age on ML balance control using a ML balance assessment task (MELBA), which consists of tracking predictable and unpredictable visual targets with the body's CoM. These tasks were used to assess the responsiveness of the balance control system, expressed in terms of control bandwidth. We found a significant effect of age for all descriptors of control bandwidth even though our older participants scored near maximum values on all clinical balance tests. The gradual increase in phase shift and decrease in gain with increasing frequency observed in both groups and for both targets (Figure 5.2) shows that MELBA tasks are challenging enough to avoid ceiling effects. In contrast, no age effect on ML-CoM postural sway and CoP postural sway during quiet standing were found. The reliability of descriptors of ML balance control bandwidth was also studied. Although small but significant learning effects between sessions, were present, reliability of the descriptors was fair to excellent with ICCs ranging from 0.57 to 0.95.

Although widely used, the evidence for the association of posturographic measures and fall risk in the elderly is inconclusive[148] and age-related changes in postural sway are controversial [158]. In the present study we found overall no age effect and only a trend towards a lower CoP-sway velocity in the ML direction in the older adults. While lower velocity would conventionally be interpreted as reflecting better balance performance, this may be attributed to a reduced exploratory behavior in the older adults, affecting functional variability hence stability [269]. Conversely, it may also reflect the reduced control bandwidth in our older participants revealed by MELBA.

Clinical measures of balance and mobility for older adults were used in the present study, to characterize the subject sample. The near-maximum scores obtained corroborate the ceiling effects reported in community-dwelling older adults [242] and underline that our sample was relatively healthy and fit. For all subjects tested, scores fell within the maximum ranges of the tests. On average, subjects were predicted to have a low risk of falling (QS = 0-1 points [263], BBS = 43-56 points [264]; MB = 19-28 points [265] and TUG and even DTUG < 13.5s [267]) , no balance impairments (POMA-B = 14-16 points) [266] and no risk of developing a future disability (SPPB = 10-12 points) [200]. The clinical tests used in this study, are thus not sensitive to subtle impairments of balance that the ML-CoM tracking tasks revealed.

Different factors may account for the lower control bandwidth observed in the older adults. The gluteus medius muscles are strongly involved in ML weight-shifting tasks [270]. When target frequency increases, faster changes in hip torques are required, which could be limited by the rate of force development of the hip abductors [271] possibly due to a selective atrophy of type-II (fast-twitch) fibers [272] and due to a reduced number of fast motor units [65]. Furthermore, tendons become more compliant with age, which can further delay force transmission and thus slow down ML balance responses [17]. It is also plausible that an increased co-activation of antagonist muscles acting in the frontal plane during the tracking tasks may hamper CoM displacement in the ML direction [112], as increased co-activation coinciding with greater stiffness and damping during ML perturbations was found in older adults [220].

In addition to changes at the effector level, impairments of the visual, vestibular, proprioceptive and somatosensory systems may affect balance control. Even though ML-CoM tracking tasks are based on, visual inputs that direct voluntary movements resulting in ML-CoM displacements, accurate online information of CoM position and velocity is needed for execution of accurate motor outputs. Deterioration of the somatosensory system due to aging may provide less accurate proprioceptive information into the balance control system [15]. Proprioceptive impairments due to aging at the hip joint have been reported [54] and may contribute to reducing ML balance control in the older adults. In addition, to proprioceptive information cutaneous plantar receptors and the vestibular organ are involved in providing sensory information into the balance control system even in the presence of explicit visual feedback on CoM movement [273]. Increased perception thresholds of cutaneous plantar receptors with aging have been reported [27] and have been associated with fall risk [263]. Also a reduced function of the vestibular system has been observed with aging [27]. The relevance of this impairment was questioned, because it was not associated with balance impairment as assessed with the POMA [27], but this may be explained by this scale not being sufficiently sensitive, as shown by the results of our study. Effects of decreased vestibular function may be more pronounced when balance is assessed with MELBA, since faster and higher amplitude body movements are made, which would rely more on vestibular information than small-amplitude and slow movements [274].

Multisensory integration is the process by which information arising from different sensory modalities is simultaneously collected [275]. Parallel weighting of sensory inputs occurs in order to control balance according to the demands imposed for a given task. For instance, impairment or absence of a sensory modality causes an up-weighting of other more reliable sources [21]. It has been proposed that the ability to re-weight sensory information as well as to perform parallel cognitive tasks is affected by aging [18,90]. Inability to properly weight sensory information and altered sensorimotor integration[228] might therefore partially explain the lower balance performance in our older adults. This is in line with previous studies that reported increased processing delays during visuomotor tasks with stepping responses [276]. Similarly, slow reactions during stepping responses have been observed in fallers who exhibited longer gluteus medius onset times [156].

Comparisons between predictable and unpredictable ML-CoM tracking tasks showed a smaller phase shift and higher gain when tracking the predictable target. This may indicate more involvement of cognitive components and more reliance upon feedback mechanisms when performing the unpredictable task [225]. Dual-tasks, used to determine the relationship between cognition and balance and balance-recovery, have shown a decreased balance performance in older adults [18]. This cognition-balance interference could be expected to cause a lower performance in the older adults, especially when tracking the unpredictable target. However, we did not find an interaction of age and task suggesting that other neuro-musculoskeletal factors, as those mentioned above, are more likely to affect ML balance performance than the decline of cognitive resources in the healthy older adults.

Although significant between-sessions differences were found, the ICC values for ML-CoM tracking performance descriptors show these to be reliable measures. All cut-off frequency descriptors (f_{PS} and f_G) had excellent reliability also in the older adults. This indicates that the bandwidth at which performance is above the thresholds (PS>-90° and G>0.5) highly correlates over sessions. The somewhat lower ICC and higher mean values for PS_{mean} and G_{mean} indicate that, within this bandwidth, performance is more variable, especially for PS and for the predictable target. Compared to the previous version of MELBA, in which CoP instead of CoM feedback was used, reliability was better in the present study, especially for the unpredictable target [225]. This may be due to the fact that ML-CoP tracking is less constrained and could allow different motor strategies, which may vary across trials and between-sessions.

The occurrence of learning effects (except for fPS) between, but not within sessions, has previously been interpreted as dissociation between the ongoing learning process and the adaptation after exposure to a novel task [277]. The results partly support the premise that visuomotor processing delay can be improved by training [278]. Although no interaction effects of session-by-age were found, differences in the average ML-CoM tracking performance between the first and second sessions were larger in the younger subjects for all descriptors except PSmean, for which improvements were larger in the older adults in both tracking tasks. Overall this indicates that also older adults are able to improve ML balance through training. However, correlations with daily-life ML balance performance using accelerometers should be assessed to explore the relevance of such training effects.

MELBA tasks aim to assess weight-shifting ability, which has been found to be deteriorated and associated to falls in older adults [6]. Performance on the predictable ML-CoM tracking may indicate maximal capacities within the requirements of the task, whereas performance on the unpredictable ML-CoM tracking can give insights into the sensorimotor integration in a more reactive manner [225]. The later may be more associated to stressing situations as those observed when internal or external perturbations are applied. Although the tracking tasks imposed do not simulate daily-life dynamic balance demands, MELBA challenges mediolateral balance control to one's maximal capacities, thereby yielding highly sensitive outcomes. Further longitudinal research needs, however, to assess the predictive value of ML balance performance on MELBA for fall risk. Finally, the utilization of less expensive and more user-friendly motion capture systems should be explored to simplify MELBA's setup to make it more clinically available.

Conclusions

In conclusion, the ability to accurately track predictable and unpredictable targets deteriorates with age. This indicates a deterioration of ML balance in apparently healthy older adults. MELBA appears to be a sensitive and reliable tool to assess ML balance performance in younger and community-dwelling older adults.

VI

Mediolateral balance and gait stability in older adults

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Abstract

Falls are a major problem in aging societies, for which early detection of balance impairment is crucial to identify individuals who may benefit from preventive interventions. Since mediolateral balance deteriorates with aging, we proposed a mediolateral balance assessment (MELBA) tool that uses a CoM-tracking task of predictable sinusoidal and unpredictable multisine targets. This method has shown to be reliable and sensitive to aging effect, however, it is not known whether it can predict performance on common daily-life tasks such as walking. This study aimed to determine whether MELBA is an ecologically valid tool by correlating its outputs with a measure of mediolateral gait stability known to be predictive of falls. Nineteen community-dwelling older adults (72±5 years) tracked predictable and unpredictable target displacements at increasing frequencies with their CoM by shifting their weight sideward. Response delay (phase-shift) and amplitude difference (gain) between the CoM and target in the frequency domain were used to quantify performance. To assess gait stability, the local divergence exponent was calculated using mediolateral accelerations with an inertial sensor when walking on a treadmill (LDE_{TR}) and in daily-life (LDE_{DL}) for one week. Pearson product-moment correlation analyses were performed to determine correlations between performance on MELBA tasks and LDE. Results show that phase-shift bandwidth for the predictable target (range above -90°) was significantly correlated with LDE_{TR} whereas phase-shift bandwidth for the unpredictable target was significantly correlated with LDE_{DL}. In conclusion MELBA is an ecologically valid tool for mediolateral balance assessment in community-dwelling older adults who exhibit subtle balance impairments.

Introduction

Falls have a high incidence in healthy elderly, with 30% of people over 65 falling at least once every year, and falls are even more common among elderly with chronic diseases and disabilities [7]. This poses a major health problem for our aging society in which above 15% of the population worldwide will be over 65 years old by 2050 [1]. Most older people exhibit some degree of balance impairment, which can increase the risk of falling [27]. Therefore detecting balance impairments at early stages in this population is crucial to identify subjects at risk of falling and ultimately of paramount importance for healthy aging.

Balance impairment and its association to fall risk have been studied using clinical and laboratory measures of balance control. Several measures of postural sway (i.e. spontaneous sway of the centre of pressure) have shown that impairment of balance in the mediolateral (ML) direction is predictive of falls [151,156]. Unfortunately, most of the current clinical balance tests do not emphasize ML balance capacities and were shown to exhibit ceiling effects. In line with this, Pardasaney and co-workers (2013) suggested that for the community-dwelling older adults, new balance assessment tools should be of greater complexity to improve sensitivity [128].

In this context, we recently proposed a ML balance assessment tool (MELBA) in which subjects track a visually presented target with ML movements of their centre of mass (CoM) [256]. MELBA was shown to be reliable and sensitive to subtle balance impairments in healthy elderly not detected by conventional posturography and clinical measures of balance [256]. Responsiveness (bandwidth) of the balance control system is assessed in terms of the response delay (phase-shift) and amplitude difference (gain) between the CoM and the target along predictable and unpredictable ML trajectories. Impairments of ML balance control likely affect gait, which is the activity during which most falls occur [9,10]. However, the association between ML balance control, as assessed with MELBA, and stability of gait is as yet unknown.

Gait stability has been quantified using the maximum Lyapunov exponent, or more appropriately the local divergence exponent (LDE)[180,279]. The LDE quantifies the sensitivity of the gait kinematics to continuous small perturbations present due to external perturbations and neuromuscular noise with greater (positive) values indicating “less stable” kinematics [280]. The LDE has been suggested to be the most suitable measure of gait stability available at present [281]. Estimates of the LDE of gait kinematics obtained during walking on a treadmill and during walking in daily-life are both predictive of fall risk [179,180,282]. Although both walking contexts assess physical capacities, daily-life walking may also include behavioural and environmental determinants of fall risk [283]. Furthermore, the LDE has been shown to be sensitive to induced impairments of balance through galvanic stimulation of vestibular afferents [284] and through external mechanical perturbations [285].

Therefore, we hypothesized that measures of balance control obtained with MELBA are associated with measures of ML gait stability in walking on a treadmill and during daily-life. Such associations would demonstrate MELBA’s predictive ability regarding gait stability and hence its ecological validity.

Methodology

Participants

Nineteen healthy older adults (7 women and 12 men, age: 72±5 years; height: 1.73±.09 m; weight: 76.6±15 kg) participated in this study. Participants were excluded if they presented any musculoskeletal or neurological condition or used medications that could affect balance. Participants had mini mental state examination scores ≥25 out of 30 [286] and clinical balance assessment that revealed maximum or close to the maximum scores above the cut-off scores for the highest category defined for each test [256].

This study was approved by the Ethical Committee of the Faculty of Human Movement Sciences, VU University (2011-48M) and the Medical Ethical Committee of the VU University Medical Center Amsterdam (2010/290), in accordance with the ethical standards of the declaration of Helsinki. All participants were informed of the experimental procedures and signed informed consent prior to the experiment.

Task and Procedure

MELBA – mediolateral balance assessment

Each participant performed a series of ML-CoM tracking tasks, while standing barefoot and with the arms crossed in a quiet and low-intensity lit room (for details of the set-up refer to Chapter V). Body CoM was calculated with a 9-markers frontal plane model (forehead, shoulder, anterior-superior iliac spines, knees and ankles) tracked with an Optotrak Certus system (NDI, Waterloo, Ontario, Canada). Gender specific CoM calculations were performed using scaling of anthropometric data and inertial parameters described by de Leva [215]. D-flow 3.10.0 software (Motek Medical, Amsterdam, The Netherlands) was used to produce target signals as well as to record (60 samples/s) and display target and CoM data on a screen 2.5 m in front of the participant. ML-CoM tracking consisted of tracking a predictable and an unpredictable target signal using the ML displacement of the CoM projected on the screen. The target signal and CoM were represented by white and red spheres of 11 and 9 cm diameter, respectively. Examples of the two target signals are depicted in Figure 5.1

The *predictable* target signal was constructed using 2 blocks of 20 seconds, 1 block of 10 seconds and 17 blocks of 5 seconds, each composed by one sine wave, which increased in frequency from 0.1 to 2.0 Hz in steps of 0.1 Hz. This information was enhanced using a metronome synchronized with the maximum displacement of the target to increase sensory input abundance. The total duration for this target signal was 135 seconds.

The *unpredictable* target signal was constructed using 15 blocks composed by the sum of 6 consecutive sine waves separated by 0.1 Hz. A pseudorandom phase-shift between sine waves between -1 to 1 period was introduced in order to avoid predictability. After each block the lowest frequency, which started at 0.1 Hz, was increased by 0.1 Hz until it reached 1.5 Hz. Duration was 40s for block 1, 20s for block 2, 10s for block 3, 8s for blocks 4 and 5, 6s for blocks 6 and 7, and 4 seconds for blocks 8 to 15. Duration of the blocks was chosen to obtain a minimum of 2 cycles per frequency. The total duration for this target signal was 132 seconds.

Each participant performed 6 ML-CoM tracking trials: 3 with the predictable and 3 with the unpredictable target. Before performing the test, one practice trial was allowed for each of the conditions. Trials were performed with at least with 1 minute of rest in between. Stance width was standardized by setting the heel distance to 11% of body height at a fixed 14° angle between the feet (Figure 5.1). Target maximum side-to-side displacement for both conditions was normalized for each subject at 50% of stance width. On average, the participants stood on the force plate with 19.0x±1.0 cm distance between heels, which determined a maximum target displacement of 9.5x±0.5 cm.

ML gait stability

Accelerations (3D) in the ML direction were recorded using an inertial sensor (Dynaport Hybrid, McRoberts, The Hague, The Netherlands) placed at sacrum with an elastic band while walking on a treadmill at a fixed 1.2 m/s steady-state speed for 5 minutes. For daily life gait ML stability, accelerations at the sacrum level were recorded during one week with a tri-axial accelerometer (DynaPort MoveMonitor, McRoberts, The Hague, The Netherlands). Participants were instructed to wear this accelerometer at all times, except during activities that could cause damage to the instrument due to contact with water (e.g. showering). The median of estimates of separate walking episodes was used for further analysis [283].

Data analysis

MELBA – mediolateral balance assessment

All data analysis was performed using custom-made software in Matlab R2011a (Mathworks, Natick MA, USA). Balance performance over the frequency ranges in the target signal was described by the gain of the linear constant coefficient transfer function between CoM and target signal from which phase-shift (PS) and gain (G) and coherence (Coh) were calculated. A detailed explanation of the method can be found elsewhere [225]. Perfect tracking performance implies PS = 0^{o} and G = 1 over all frequencies comprised in the target signal. Coh was used to corroborate the assumption of input (target)/output (CoM) linearity and therewith the validity of estimates of PS and G. Perfect linearity yields Coh = 1 over all frequencies comprising the target signal.

To characterize balance performance, 4 descriptors were calculated. First, the values at which PS dropped below 90 degrees and G dropped below 0.5 were determined as the cutoff frequencies (coined f_{PS} and f_G, respectively). Second, PS_{mean} and G_{mean} were computed as the averages of the G and PS values within the bandwidths determined by f_{PS} and f_G, respectively.

ML Gait stability

Treadmill. The Local Divergence Exponent in the ML direction (LDE_{TR}) was calculated using the method described by Wolf et al. [287] over the whole period of 5 minutes; a subsequent normalization to stride time was performed [288,289]. A full description of the calculations is presented elsewhere [288].

Daily Life. LDE in the ML direction (LDE_{DL}) was calculated using the median over multiple non-overlapping 10 seconds windows of walking episodes. The same normalization to stride time used above was performed. All analyses were performed using custom-made Matlab functions (R2011a, Natick MA, USA).

Statistical Analysis

A univariate ANOVA was performed to determine differences between predictable and unpredictable CoM-tracking performance as well differences between walking on a treadmill (LDE_{TR}) and during daily life (LDE_{DL}).Person product-moment correlation analyses were performed to determine correlations between MELBA descriptors (f_{PS}, PS_{mean}, f_G and G_{mean}) for both targets and LDE_{DL} and LDE_{TR}. For all analyses significance level was set at $p < .05$. Statistical analyses were performed using IBM SPSS (Statistics 21).

Results

Overall, performance on the predictable CoM tracking task was significantly ($p < .01$) better than on the unpredictable with PS values closer to 0 and G values closer to 1 (Figure 6.1). Control bandwidth was wider when tracking the predictable target, with higher f_{PS} and f_G ($p < .01$) and higher PS_{mean} and G_{mean} within these bandwidths ($p < .01$)(Table 6.1). Mean LDE values were significantly lower (more stable; $p < .01$) when walking on the treadmill than during daily life.

Results for all linear regression analyses are presented in table 6.2 whereas figure 6.2 shows scatter-plots for the significant correlations found. Linear regression analyses revealed that f_{PS} for the predictable target was significantly correlated to LDE_{TR} (*r*= -.48, *p*=.04) whereas f_{PS} for the unpredictable target was significantly correlated to LDE_{DL} (*r*= -.57, *p*=.01). Other MELBA descriptors for both targets did not exhibit significant correlations either with LDE_{DL} nor with LDE_{TR}.

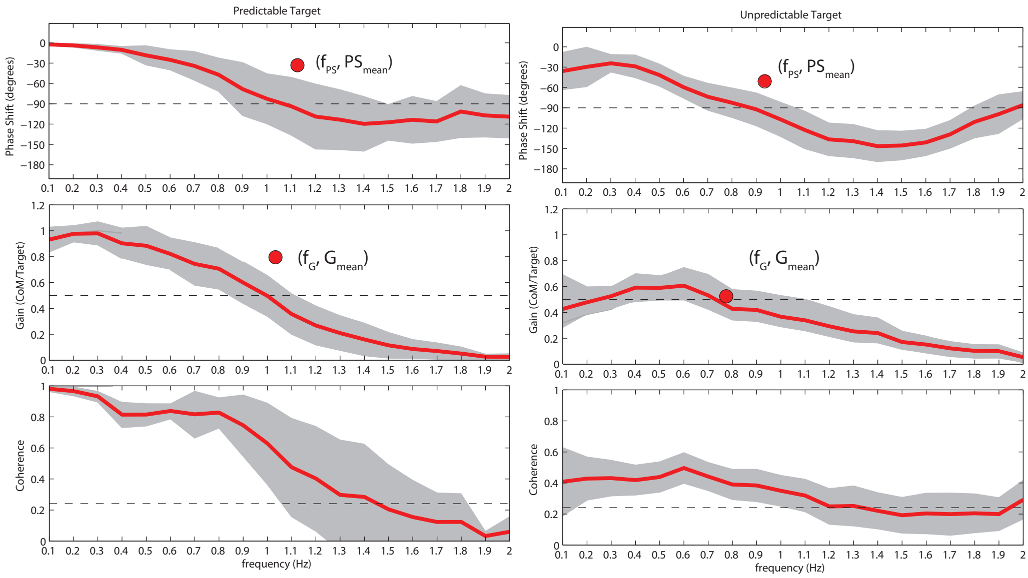


Figure 6.1. Averaged curves (± sd) for phase shift (top panel), gain (mid panel) and coherence (bottom panel) measures using both, predictable target (left) and unpredictable (right) targets. Grey shading indicates the ± sd for all subjects and for all trials. Circular markers inserted in the plots indicate means for performance descriptors.

Table 6.1. Descriptive statistics for all MELBA descriptors (±sd) for both targets and ML gait stability measures (LDE) on both settings (treadmill and daily-life) are presented on the top and bottom part of the table, respectively. Right side of the table presents 95% confidence interval ranges and significant differences (*p*-values) when comparing performance descriptors between targets and measures of gait stability between settings.

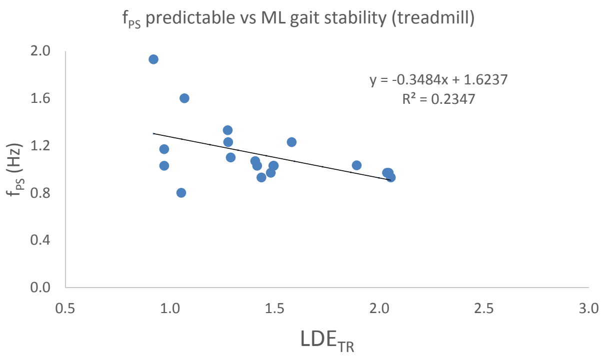
			mean	sd	95% confidence		*p*
Unpredictable	f_{PS}	(Hz)	.96	.19	0.86	1.05	
	PS_{mean}	(º)	-49.91	6.54	-53.04	-46.40	
	f_G	(Hz)	.80	.21	0.70	0.91	
	G_{mean}		.53	.09	0.48	0.57	**<.01**
Predictable	f_{PS}	(Hz)	1.13	.26	0.99	1.26	
	PS_{mean}	(º)	-32.81	5.32	-35.65	-30.24	
	f_G	(Hz)	1.04	.14	0.97	1.11	
	G_{mean}		.79	.05	0.77	0.82	
Stability	LDE_{TR}		1.43	.36	1.25	1.62	**<.01**
	LDE_{DL}		2.01	.39	1.82	2.21	

Discussion

Early detection of balance impairments is crucial to identify older adults at risk of falls and further impairments. Therefore, sensitivity to subtle changes in balance is imperative for assessment tools [128]. Besides sufficiently sensitive, a method must be ecologically valid and consider the main factors that challenge balance in daily-life activities. Since measures of gait stability appear to be predictive of falls, MELBA's association with gait stability during treadmill and daily-life demonstrates its ecological validity. Significant associations between LDE_{TR} and f_{PS} (control bandwidth) for the predictable target and LDE_{DL} and f_{PS} for the unpredictable target were found but not between LDE_{DL} and f_{PS} for the predictable and LDE_{TR} nor between LDE_{TR} and f_{PS} for the unpredictable.

Table 6.2. Results for the Pearson product-moment correlation analyses performed between MELBA performance descriptors for both targets and ML gait stability measures (LDE). Left side of the table shows *r*- and *p*-values for the treadmill walking (LDE_{TR}) whereas right side presents test statistics for the daily-life condition (LDE_{DL}). Significant correlations (*p*<.05) are highlighted in bold.

		LDE_{TR}		LDE_{DL}	
		r	*p*	*r*	*p*
Predictable	f_{PS}	**-.48**	**.04**	-.40	.10
	PS_{mean}	-.40	.09	-.27	.27
	f_G	-.31	.20	-.12	.64
	G_{mean}	-.29	.23	-.35	.16
Unpredictable	f_{PS}	-.46	.05	**-.57**	**.01**
	PS_{mean}	-.19	.43	-.25	.33
	f_G	-.15	.54	-.05	.84
	G_{mean}	-.08	.73	.06	.82



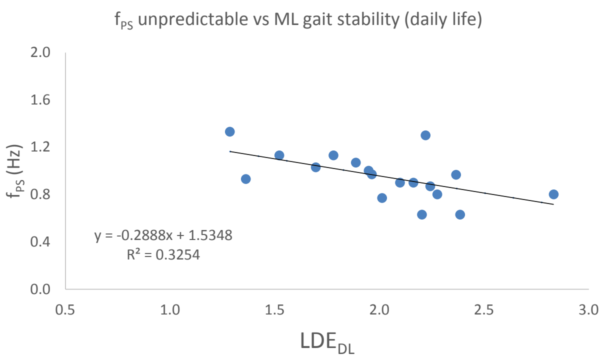


Figure 6.2. Scatter-plots showing significant (*p*<.05) correlations found between f_{PS} for the predictable target and LDE_{TR} (top panel) as well as f_{PS} for the unpredictable target and LDE_{DL} (bottom panel). Regression equations as well as R^2 values are also presented within the figure.

A study comparing gait stability in daily-life walking and when on a treadmill found that the latter was more symmetric, less variable and more stable [283]. Since, in this experiment, unexpected challenges to the balance control did not occur during treadmill walking, stability in this task was likely determined mainly by physical capacities. MELBA's predictable task also assesses this aspect of balance, which may explain the association between LDE_{TR} and f_{PS} for the predictable target [225]. During treadmill walking as well as during predictable CoM-tracking, a fixed weight-shifting pattern is followed. However, whereas for treadmill walking this pattern is constant, in MELBA, physical capacities are progressively further challenged by increasing the frequency of the target to be tracked yet maintaining the amplitude of the ML displacement.

The significant associations between LDE_{TR} and f_{PS} (control bandwidth) for the predictable target and LDE_{DL} and f_{PS} for the unpredictable target may indicate that similar ML balance resources are utilized during CoM-tracking tasks and walking in daily life. The unpredictable nature of the context of daily-life walking, where environmental challenges such as uneven terrain or potential collisions with other people, require adjustments of the gait pattern may explain this association. Gait adjustments likely require fast sensory integration to control weight-shifts similar to those required during the unpredictable CoM-tracking task. It has been previously reported that incorrect weight-shifting accounts for 41% of falls in residential care facilities, which mainly occurred during walking [6]. Although we assessed community-dwelling older adults, MELBA's sensitivity to age [256] indicates that weight-shifting may also be an early sign of balance deterioration in healthy elderly.

The characterization of gait stability using accelerometers during treadmill and daily-life walking has been shown to predict falls in the elderly population, hence offering an ecologically valid measure of balance performance [179,180,282]. However, since stability-threatening events do not necessarily occur on a regular basis, these measures may not reflect one's ability to cope with strong balance threats [281]. Challenging the balance system to its maximal capacities is crucial to determine subtle impairments that may hamper responses to external perturbations, especially in able-bodied older adults. In this respect, MELBA has shown to be challenging enough so as to observe CoM-tracking performance consistently dropping below PS and G thresholds even in healthy young subjects [256].

It has been reported that impairments of different systems contributing to balance control are affected by aging [27,54]. This is likely to affect performance during CoM-tracking tasks as well as stability during walking; however, sensory re-weighting and changes in motor strategies may occur to compensate for sensorimotor deficits and avoid instability during both, MELBA and walking. Balance assessment measures should, therefore, aim to maximize the contribution of each system when assessing an older person's maximal capacities. When compared to clinical and posturographic measures, MELBA has shown to be more sensitive to aging and hence likely demands each balance sub-system's contribution to a greater extent [256]. In addition, the use of visual feedback is not likely to mask the impairment of other sensory systems [273].

While LDE_{DL} obtained over a full week [283] and MELBA performance descriptors [256] have been shown to have good reliability, LDE_{TR} over a single session has been shown to be less reliable [290]. This limited reliability may have affected associations between LDE_{TR} and MELBA descriptors. Although other measures of gait stability have also been shown to predict falls in the elderly [283], we only focus on the ML direction, since compelling evidence points to balance on this plane as the most affected by aging when standing and walking [6,156,168,282,291]. Further studies should explore whether the combination of MELBA and gait stability measures has added value for the prediction of falls in the older adults in prospective studies.

Conclusion

Significant correlations between mediolateral stability during treadmill and daily-life walking and ML balance as determined with MELBA, support the ecological validity of this tool for ML balance assessment in community-dwelling older adults, who exhibit subtle balance impairments.

VII

Epilogue

Introduction

The aim of this thesis was to develop and test a method for quantification of mediolateral balance in community-dwelling older adults. In the first study described in **Chapter II** of this thesis, possible masking of balance impairment due to up-weighting of visual inputs when using explicit visual feedback on performance was studied using sensory manipulations. In **Chapter III**, reliability of a mediolateral balance assessment method (MELBA) that utilizes CoP tracking of sinusoidal predictable and unpredictable targets was studied in a group of young adults. In **Chapter IV,** the advantages of using CoM instead of CoP feedback are investigated. **Chapter V** presents a modified version of MELBA that utilizes CoM instead of CoP tracking over similar predictable and unpredictable targets. Besides reliability, the effect of age on performance and comparisons to clinical and conventional posturography measures were studied. **Chapter VI** of this thesis presents a study on the ecological validity of MELBA by correlating outcomes to measures of mediolateral gait stability on a treadmill and during daily-life.

Enhancing visual information…. yet not too much.

There is an emerging interest in using commercially available gaming technologies as means of assessment and training in subclinical an clinical populations, for which the term "serious-games" (SG) has been coined [292]. The vast majority of the applications utilized by SG heavily rely on visual feedback to engage with the controller (i.e. patient) [251]. Since it is known that flexibility of the balance control system allows for up-weighting of more reliable sensory information [21,81,82], impairment of other sensory sources may be masked and eventually overlooked, when assessment includes visual feedback which may increase the gain of visual inputs. In **Chapter II**, an experiment in which vestibular, proprioceptive and somatosensory systems were disturbed while subjects were presented with (explicit) visual feedback on ML-CoM sway was conducted to determine whether erroneous inputs due to sensory manipulations can be completely neglected by the balance control system in the presence of visual feedback. Results showed that the effect of sensory manipulations is not completely effaced by the presence of visual feedback. However, healthy young volunteers were measured in this study for whom effects of sensory manipulation may differ from older adults or pathological populations, who rely more on visual input [89,90]. The latter groups may exhibit an increased ability to minimize the effect of sensory manipulations under visual feedback on performance. The extent to which this compensatory sensory weighting mechanism may be advantageous to maintain stability in older adults is yet to be explored. However, it seems unlikely that effects of impairments of other sensory modalities would be completely effaced given impairments of sensory integration with aging [57,92]. Furthermore, it has been shown that not all healthy older adults take advantage of CoP visual feedback to increase stability [222]. It is also possible that up-weighting of a sensory modality may have a fixed maximum gain associated to internal adaptations of the balance control system due to age-related multisensory deterioration which ultimately affects the dynamics of sensory integration.

The results on Chapter II support to the use of visual feedback for assessment of balance, specifically in the mediolateral directions. The use of visual feedback, as previously mentioned, is implicit in most SG, and our results do not rule out that increased visual system reliance accounts for balance improvements observed after training [293]. Although balance improvements do not appear to translate into gait speed and TUG performance in community dwelling older adults [294], improvements on the BBS score have been seen in frail elderly [295] as well as a reduced incidence of falls [296]. This may indicate that even if training with visual feedback increases weighting of visual information, this has beneficial effects on daily-life balance performance [212].

A simple CoP tracking task

In **Chapter III,** the first version of MELBA using a visual CoP-tracking task was introduced. This task allowed calculation of performance descriptors reflecting the maximum performance of the subject. Performance indicators of CoP-tracking of predictable and unpredictable targets exhibited good to excellent between-sessions reliability in young adults. Whereas performance on the predictable target may reflect physical capacities and feed-forward balance control, the unpredictable may demand reactive control to a greater extent, requiring a more complex integration of information. Performance descriptors were calculated based the individual's maximum performance during a trial. Although beneficial in terms of individualized assessment when aiming for a customized intervention, it does not allow standardization of the test, so as to determine cut-off scores for classification of individuals at risk of balance impairment and/or falls. This would require absolute thresholds on performance. Perhaps, this individualized approach can serve for the therapeutic purpose of determining a subject's progression over time whereas the second may serve for identifying subjects at risk of falling due to ML balance impairments.

As with ageing, in some neurological disorders such as Parkinson's disease and stroke incorrect weight-shifting is associated with an increased number of falls. CoP-tracking tasks using MELBA may better quantify these impairments as well as eventual treatment effects. Since not originally conceived for this population, frequency content as well as side-by-side amplitude of CoP movement may need to be adapted to make it more suitable for clinical populations. This may not only avoid floor effects, a concern when assessing mobility impaired subjects, but also suggests potential as a customizable training tool for which balance demands can be set according to progression. Despite the fact that forceplates are not widely available in clinical settings, current gaming technology such as the Wii balance board may offer platform at which MELBA can be implemented for assessment [252,297,298]. This is of particular interest since more sensitive assessment applications, alternative to Wii Fit, which is not effective in determining balance status, have been suggested [299,300]. The use of scores in common Wii games has been reported to discriminate between older adults with and without history of falls [251]. Although this scored-based assessment was conducted when sitting, discriminative ability of those games may indicate that the reactive control demanded in these games may contribute to prevention of falls [251].

A more intuitive CoM tracking task

In the first version of MELBA that used CoP feedback, balance performance may be overestimated when the inverted pendulum assumption does not hold at high frequencies and at which strategies in the sagittal plane (i.e. on-site stepping) or transverse plane (i.e. trunk and pelvis rotations) even though subjects are instructed to shift weight from left to right as if they were a "stick". In **Chapter IV** , the benefits of using a CoM instead of CoP tracking tasks are analysed. This study was aimed to determine whether CoP ML displacements elicit consistent CoM ML displacements, as well as to determine the implications of using an explicit CoM tracking task. Despite the greater computational and measurement effort involved in calculating the CoM, tracking tasks using CoM may be preferable, since they elicit a wider variety of kinematic strategies, which may allow for more comprehensive exploration of motor control possibilities in realizing weight shifting . Although calculations of CoM require greater computing power than estimations of CoP, it is though that as the accuracy of simple motion capture systems improves (e.g. Kinect[257]), both tracking tasks will not differ in terms of costs.

An important advantage of using the CoM, is the fact that this is the controlled variable in balance control [253] which is likely to elicit more intuitive balance control responses. Besides, a shift from ankle to hip and trunk kinematic strategies was observed during CoM tracking which may serve to explore the age-related distal-to-proximal shift in the locus of function as observed during gait [255]. However, the contribution of musculature related to stabilization and CoM ML displacements during MELBA is yet to be explored by using electromyography on the main frontal plane actuators which are thought to be in/eversors of the ankle, hip ad/abductors and trunk muscles. Furthermore, it is possible that the activation pattern during the CoM tracking tasks is affected by aging, which will be reflected in leg and trunk kinematics. Determining changes in CoM tracking strategies may help in identifying and targeting preventive or rehabilitative interventions to preserve optimum balance control.

A revised version of MELBA.

In **Chapter V** a newer version of MELBA using CoM feedback on performance is described. CoM-tracking has the advantage of being more meaningful and intuitive because the CoM is the variable to be controlled when dealing with internal and external balance perturbations [253]. In this study, both performance descriptors for the MELBA tasks exhibited good to excellent between-sessions reliability and most importantly were sensitive to age effects. This contrasted with results obtained from posturographic measures that did not discriminate between young and older adults, as in previous studies [149-152] and for which this method has been severely criticized [132]. Possibly posturography is not challenging enough to determine subtle impairment of balance control in the able-bodied older population. In this regards, MELBA offers a task in which the balance system is challenged to its maximum capacity under feedforward (predictable task) and reactive (unpredictable task) conditions.

In the revised MELBA version, we used fixed thresholds to determine performance descriptors, which allowed comparing performance of young and older adults. The current threshold for phase-shift (90 degrees) was chosen considering that a motor reaction after .25 of a cycle means that CoM and target will have an opposite direction during half a cycle, implying that for frequencies above 1.4 Hz a 90 degrees delay will demand reaction times above those exhibited by young adults in clinical tests (>182 msec)[138]. On the other hand, a threshold of 0.5 for the gain will demand around 5 cm ML CoM displacement when tracking a single sine wave (average ML displacement in MELBA is ≈10 cm). This displacement is similar to the displacement exhibited during walking at relatively normal speed (≈1.2 m/sec) and is average between walking at 1.0 m/sec (slow) and 1.6m/sec (fast) [301]. However, selected thresholds can be modified (lowered or heightened) depending on future research that may determine more sensitive thresholds, for instance, to predict falls.

Most common clinical tools for balance assessment have been validated on subjects with pathologies affecting their mobility, or in fragile older adults. In our study, all healthy older participants scored within the range defined for the maximum performance supporting the claim that in contrast with such clinical methods MELBA does not suffer from ceiling effects when used in able-bodied older adults [127,128]. Several previously made suggestions for better test of balance assessment were incorporated in MELBA. However, an important aspect that was not incorporated is the use of external perturbations and sensory manipulations [132,145,163]. Although they may provide insightful information on reactive components of balance that demand faster sensorimotor integration, the added value of such perturbations tasks when determining subtle balance impairment is yet unknown. In addition, implementation of controllable perturbations may involve higher costs and safety issues, which constrain use in clinical settings, especially when using it as a screening tool.

Since CoM-tracking tasks involve the use of motion capture systems for online computation, this may limit its use as an easy implementation tool. However, as well as for force plates, gaming technology offers inexpensive platforms such as the Microsoft Kinect in which a MELBA can potentially be implemented [257]. This technology may ease the access not only for MELBA assessment, but also for other potential motion-performance based assessments of balance and for fall prevention training [302].

MELBA to the test: ecological validity

In **Chapter V**, we concluded that MELBA using CoM as feedback on performance is a reliable and sensitive tool for mediolateral balance assessment in able-bodied older adults. It was then questioned, how does this performance relate to balance control in more ecological tasks? In **Chapter VI,** performance on MELBA was correlated to ML gait stability during laboratory (treadmill at fixed speed) and daily–life walking (one week). Stability was quantified using the local divergence exponent (LDE), which for gait in daily-life was significantly correlated with the phase-shift cut-off frequency (f_{PS}) for the unpredictable target and for gait on the treadmill to the same parameter for the predictable target. The phase-shift is a measure of the response delay and can be associated to the reactive components of balance control. The moderate correlations values found may indicate that MELBA tasks are more challenging than the control of weight shifts that normally occur when walking, with smaller ML CoM displacement [301] and higher frequency content [188]. Since in healthy individuals stability-threatening situations may occur only when coping with high magnitude disturbances, MELBA could be a more sensitive to early balance impairments than gait stability measures.

It has been shown that older adults have an increased energetic cost when walking compared to young adults, possibly due to an increased co-activation of agonist and antagonist muscles [303]. This increased energetic cost has been also associated to impairments of mediolateral balance [304]. Possibly, increased muscle co-activation in the elderly is functionally used to increase muscle spindle feedback gain when other sensory sources such as somatosensory inputs are deteriorated [35]. Moreover, increased co-activation could increase the bandwidth of the control system, by reducing electromechanical delays [305]. This compensatory mechanism may be of specific relevance during weight transfer and single stance in gait and reflected in a smaller phase-shift and gain closer to 1 when exhibited by the older adults.

Although this study provided some indication of ecological validity, several other motor tasks than unperturbed gait may impose greater balance demands, such as quick direction changes or ground perturbations. To determine whether MELBA is an ecologically valid tool for other tasks demanding weight-shifting, correlational analysis comparing between MELBA descriptors and performance on such tasks could be performed (i.e. sudden ML ground perturbations). Considering that all subjects (young and older adults) dropped their performance below the thresholds set for phase-shift and gain (no ceiling effect), even stronger correlations with performance on more challenging ML balance tasks might be expected.

Further investigation on the ecological validity of MELBA should also prospectively explore its correlation with the incidence of falls experienced in the older population. If MELBA's performance indicators predict falls, this relatively simple assessment tool can be used as a screening test to identify subjects in need of preventive interventions. This would likely increase efficiency of efforts aimed to maintain quality of life and mobility in older adults.

Clinical implications: what's new?

Although the use of weight-shifting tasks with CoP feedback on performance has been studied before in young, elderly and clinical populations [223,226,306,307], MELBA is the first assessment tool that uses visual tracking tasks (CoP and CoM feedback) to quantify balance using a linear systems identification approach. The clinical implications of the results of the studies presented in this thesis can be summarized as follows:

- MELBA is a reliable tool for mediolateral balance assessment in the able-bodied population.
- It has no ceiling effect, since frequency content can be increased as well its resolution.
- It is likely not to exhibit floor effects since frequency content can be lowered and range constrained.
- Using CoM instead of COP tracking offers a more intuitive assessment method for which a broader spectrum of kinematic strategies are called into play.
- MELBA challenges weight-shifting ability, which seems to be an early indicator of balance impairment.
- It assesses proactive (predictable task) as well as reactive components (unpredictable task) of balance control.
- It is an ecologically valid tool.
- It has the potential of becoming an inexpensive computerized assessment tool once implemented using gaming technology.
- It will not require technical expertise or motion analysis skills once "normal" and "pathological" performance thresholds have been determined.
- It has the potential of becoming a relatively simple computerized clinical tool, that is easy to implement and not time-demanding.
- It is possible that MELBA-like training methods (weight-shift training) can maintain or improve balance in older adults.

Although MELBA is a promising balance assessment tool, several suggestions from previous research to develop new methods were not taken into account. Perhaps the most important one is that MELBA does not look at the balance responses under mechanical perturbations. However, maximal capacities exhibited during the test may be less risky than applying standardized mechanical perturbations even in healthy older adults. It is noteworthy, that ongoing research is looking at correlations between ML sudden perturbations and MELBA performance as well as balance responses to sinusoidal mechanical perturbations following the same pattern and amplitudes as in our current predictable and unpredictable CoM tracking tasks. This may allow determining the added value of using mechanical perturbations as part of clinical assessment in older population.

Since neurological conditions such as Parkinson's disease and stroke survivors as well as musculoskeletal conditions also affect mediolateral balance control (weight-shifting), it is possible to extend the use of MELBA to these populations. The recovery of weight-shifting abilities assessed with MELBA may indicate reduced risk of falling as well improvements after therapeutic interventions. Although promising for clinical applications, performance on MELBA should not be considered as a single tool to be used when prescribing interventions, but rather as part of a more comprehensive assessment since other factors such as fear of falling may also affect balance in clinical populations as well with aging [308].

Serious-gaming may offer an engaging alternative to conventional balance training [309,310] and fall prevention programs [311] increasing compliance with intervention programs in older adults [312], especially when aiming for long-term maintenance using regular exercise or balance training [313,314]. The use of serious-gaming has already been shown to have positive effects on balance in healthy [250] and frail [294] older adults, and also in clinical populations [315]. However, it is still not known whether benefits are greater or complementary to multidimensional [310] or specific balance training [316,317], group based exercises [318], power and strength training of legs [78] and trunk [319] or Tai-Chi [320] to improve balance and reduce the risk of falling.

Future research: Early detection of balance impairment or falls prediction?

In **Chapter II**, sensory manipulations were used to assess sensory weighting processes that may mask balance impairments when providing explicit visual feedback on performance. However, there are still several fundamental questions that should be explored in relation to the effect and extent of these manipulations. The effects of TENS as a means to disrupt somatosensory information should be further compared with more traditional methods used to remove or perturb these inputs such as feet cooling or standing on foam. Although, vibrational muscle-tendon stimulation effects have been widely reported, it is still not known what is the minimum amount of time to elicit the “lengthening illusion”. It is also unknown whether stronger effects occur when the muscle is relaxed or active and whether more spread manipulations (more than a single pair of muscles simultaneously vibrated) can further affect proprioceptive control. Other questions still to be explored are related to the function of vestibulospinal reflexes elicited by the GVS. For instance, it is still not clearly known whether these reflexes are elicited by linear, angular or both types of accelerations and how they contribute to maintain stability under threatening situations.

It is possible that weights on inputs from impaired systems may have a fixed maximum gain, hence maintaining some reliance on other sensory sources. This hypothesis can be further tested by enhancing impaired sensory channels and determine whether their weight can be further increased when compared to the non-enhanced condition when disrupting other sensory sources. An issue to be further explored is how to normalize different types of sensory manipulations to allow proportional manipulations when studying sensory weighting and sensorimotor integration mechanisms. Also further studies could explore how balance control deals with multiple sensory manipulations, to emulate multiple sensory impairment in older adults or clinical populations.

In relation to the use of MELBA as a computerized clinical assessment or screening tool for community-dwelling older adults, implementation on simpler gaming platforms, such as Kinect, should be explored to minimize the time needed for its application. This will also allow collection of larger data sets to determine reference values of performance. Consequently, it will allow identifying older adults with greater compromise of balance control or increased risk of falling that may benefit from preventive programs. Prospective studies aimed to determine MELBA’s predictive validity should be carried out to determine its predictive ability for falls and mobility impairment and disability.

It is possible there is a common path for age-related and pathological balance impairments. This calls for further investigation, first on the use of MELBA in clinical populations and second to determine the neurophysiological causes for inadequate weight-shifting. Finally, the implementation of serious-gaming in which mediolateral balance training is emphasized should be explored as a means to maintain or improve ML balance in older adults population.

Conclusion

The aim of this thesis was to develop and test a method for quantification of mediolateral balance in community-dwelling older adults. Based on the studies in this thesis it can be concluded that the use of CoP visual feedback does not mask impairment of other sensory sources, which allows its use for balance assessment. The use of CoP as visual feedback on performance during predictable sinusoidal and unpredictable tracking (MELBA) offers a reliable tool for balance assessment. To improve MELBA, CoP feedback was replaced by CoM feedback, which was again shown to be reliable in young and moreover in older adults. Furthermore, this test was demonstrated to be more sensitive to aging than clinical tools and conventional posturography measures. Additionally the ecological validity of MELBA was demonstrated by significant correlations between MELBA performance indicators and the mediolateral stability when walking on a treadmill and during daily-life. Further studies should investigate the ability of MELBA to predict falls and its feasibility for clinical implementation in larger groups of older adults.

S

Samenvatting

Summary

Resumen

Nederlands

De algemene inleiding van **Hoofdstuk I** geeft een overzicht van het belang van balanscontrole bij ouderen. Aandoeningen in de balanscontrole, vooral in zijwaartse richting, zijn een belangrijke oorzaak van vallen, die ernstige gevolgen hebben op de zelfredzaamheid en kwaliteit van leven van ouderen. Zijwaartse (mediolaterale) balanscontrole is vooral belangrijk in taken waarbij het lichaamsgewicht verplaatst wordt, zoals bij (op)staan en lopen. Afname van de functies van het sensorische en motorische systeem, maar ook van cognitief functioneren kan een negatief effect hebben op de sensomotorische integratie en dit leidt tot een afname van het vermogen om met balans verstorende situaties om te gaan. Daarom zijn testen nodig, die gevoelig genoeg zijn om een afname in de balanscontrole in een vroeg stadium te detecteren, zodat een adequate behandeling kan worden ingezet om verdere afname te voorkomen. De huidige klinische en onderzoeksmethoden zijn helaas nog niet gevoelig genoeg om kleine afname in de balanscontrole bij gezonde thuiswonende ouderen vroegtijdig op te sporen, mogelijk omdat deze testen niet uitdagend genoeg zijn. Het doel van het onderzoek in dit proefschrift was het ontwikkelen en testen van een visuele volgtaak om mediolaterale balans te meten bij gezonde, fitte ouderen.

In **Hoofdstuk II** is onderzocht of visuele informatie en expliciete feedback van de mediolaterale bewegingsuitslag van het lichaamszwaartepunt verstoring van de sensorische informatie zou kunnen maskeren. In deze studie werd bij jong volwassenen informatie uit het evenwichtsorgaan verstoord door middel van galvanische stimulatie, proprioceptie door spier-pees vibratie en somatosensorische informatie van huidsensoren door transcutane zenuwstimulatie, tijdens stilstaan met de ogen dicht, ogen open of met expliciete feedback. Uit de resultaten bleek dat de bewegingsuitslag van het lichaamszwaartepunt door de sensorische verstoringen niet verdween, wanneer expliciete visuele feedback werd gegeven. Dit geeft aan dat eventuele verstoringen of aandoeningen in sensorische systemen niet zullen worden gemaskeerd door het gebruik van visuele feedback in een volgtaak.

In **Hoofdstuk III** wordt een mediolaterale balans test (MELBA) geïntroduceerd. Bij deze test moet de proefpersoon (multi)sinusvormige, voorspelbare en onvoorspelbare visuele targets volgen op basis van feedback van het aangrijpingspunt van de grondreactiekracht (CoP). Het vermogen om de volgtaak uit te voeren wordt bepaald door de transfer functie tussen het target en het CoP waaruit de phase-shift (vertraging in de respons), de gain (verschil in amplitude) en de coherentie (mate van lineariteit) worden berekend. De uitkomst wordt bepaald op basis van de bandbreedte (de frequentie waarop de phase-shift of gain onder een grenswaarde van respectievelijk -90° of 0.5 daalt) en de gemiddelde waarden van de phase-shift en gain binnen deze bandbreedtes. Uit deze studie bleek dat MELBA een betrouwbare test is. Bovendien werden kleine, maar consistente leereffecten gemeten, wat aangeeft dat een dergelijke methode gebruikt kan worden voor training.

Aangezien tijdens balanscontrole het lichaamszwaartepunt gecontroleerd moet worden, wordt bij een volgtaak met feedback van het CoP aangenomen dat het lichaamszwaartepunt een consistente beweging vertoond. Deze aanname werd onderzocht in **Hoofdstuk IV**, waarbij de hoeveelheid verplaatsing van het lichaamszwaartepunt werd gemeten tijdens volgtaken met feedback van het CoP en het lichaamszwaartepunt en waarbij de verschillen in bewegingsstrategieën tussen deze feedback modaliteiten werden onderzocht. Het bleek dat feedback van CoP consistente bewegingsuitslag van het lichaamszwaartepunt liet zien. De uitvoering van de beweging verschilde echter tussen de twee vormen van feedback, waarbij feedback van het lichaamszwaartepunt een grotere buiging van de romp liet zien, en dat met name bij de hogere frequenties rotaties meer in de heup dan in de enkel plaatsvonden. Dit geeft aan dat feedback van het lichaamszwaartepunt de voorkeur heeft bij visuele volgtaak, ondanks extra meetbelasting.

De studie in **Hoofdstuk V** had als doel om (1) het effect van leeftijd en (2) de betrouwbaarheid te bepalen van de nieuwe versie van MELBA met feedback van het lichaamszwaartepunt. Zowel jong volwassenen als ouderen voerden de MELBA taken uit en alle jongeren en een deel van de ouderen werden nogmaals gemeten voor het bepalen van de betrouwbaarheid. Bovendien werd standaard posturografie en bij alle ouderen een aantal traditionele balanstesten afgenomen. Uit de resultaten bleek dat de bandbreedtes bij ouderen kleiner waren dan bij jongeren en dat de gemiddelde waarden van phase-shift en gain binnen die bandbreedtes ook lager waren. De testmaten bleken zeer betrouwbaar voor beide leeftijdsgroepen en vertoonden kleine, maar consistente verbeteringen tussen sessies, wat een leereffect suggereert. In tegensteling tot de MELBA uitslagen scoorden de ouderen allemaal nagenoeg maximale scores op de traditionele balanstesten en standaard posturografie vertoonde geen leeftijdseffect. Op basis van deze studie kan geconcludeerd worden dat MELBA een betrouwbare test is om mediolaterale balans in jongeren en ouderen te meten, dat het vermogen om deze test uit te voeren afneemt met leeftijd en dat MELBA gevoeliger is voor leeftijdsveranderingen in balanscontrole dan traditionele balanstesten en standaard posturografie.

De ecologische validiteit van MELBA is bestudeerd in **Hoofdstuk VI**. Hier werden de balansmaten uit MELBA gebruikt om de mediolaterale stabiliteit tijdens het lopen op een loopband en in het dagelijks leven te voorspellen bij een groep gezonde, fitte ouderen. De stabiliteit tijdens lopen werd bepaald door de local divergence exponent te berekenen op basis van mediolaterale romp versnellingen, gemeten met een versnellingsmonitor tijdens het lopen op een loopband gedurende 5 minuten of in het dagelijks leven gedurende een week. Het bleek dat de phase-shift bandbreedte van de voorspelbare volgtaak significant gecorreleerd was met de stabiliteit tijdens lopen op de loopband, terwijl de phase-shift bandbreedte van de onvoorspelbare taak gecorreleerd was met de stabiliteit in het dagelijks leven. Dit suggereert dat de uitkomst van de voorspelbare volgtaak indicatief is voor het fysieke vermogen voor mediolaterale balanscontrole in onverstoorde situaties, terwijl de uitkomst van de onvoorspelbare volgtaak indicatief is voor het vermogen om in het dagelijks leven met onverwachte balansverstoringen om te gaan. Er kan geconcludeerd worden dat MELBA een ecologisch valide test is voor gezonde ouderen met kleine balansproblemen.

Dit proefschrift beschrijft de introductie en toetsing van een nieuwe mediolaterale balanstest (MELBA). Het is aangetoond dat deze test betrouwbaar, richtingsgevoelig en uitdagend is, geen plafondeffect vertoont en gevoeliger is voor leeftijd dan conventionele balanstesten. Op basis van deze eigenschappen is het mogelijk om met deze test kleine aandoeningen in de balanscontrole van ouderen te identificeren, die voorspellend zijn voor instabiliteit in het dagelijks functioneren. In **Hoofdstuk VII** worden deze bevindingen, mogelijke beperkingen en klinische implicaties, alsmede aanwijzingen voor vervolgonderzoek bediscussieerd.

English

The general introduction in Chapter I contains an overview of the importance of maintaining balance abilities in older adults. Balance impairments, in particular in mediolateral direction, have been indicated to be one of the major causes of age-related falls that can have serious consequences for mobility independence as well as wellbeing and quality of life. Furthermore, evidence suggests that mediolateral balance is required for weight-shifting abilities and inabilities seem to play an important role in falls in older adults that occur when standing, stepping or walking. Age-related impairments of sensory and motor systems as well as cognitive impairments negatively affect the process of sensorimotor integration. This can ultimately deteriorate one's balance performance when coping with hazardous situations in daily-life. Sensitive tools are therefore required for early detection of balance impairment in the elderly as this may serve to screen individuals that can benefit from interventions to prevent balance disorders. Currently available laboratory and clinical tests of balance, however, exhibit ceiling effects when determining subtle balance deterioration in community-dwelling older adults, possibly because they are not challenging enough for the balance control system in comparison to daily-life exposure to hazardous situations. The aim of the work presented in this thesis was to develop and test a visual tracking task that can quantify mediolateral (ML) balance impairments and that is suitable for highly functioning older adults.

Chapter II presents a study aimed to determine whether visual information and explicit visual feedback on mediolateral sway of the body centre of mass (CoM) might reduce erroneous sensory inputs introduced into the balance control system in healthy young adults. This study utilized manipulations of the vestibular (galvanic stimulation), proprioceptive (muscle-tendon vibration) and somatosensory (transcutaneous nerve stimulation) systems during standing still with eyes open, eyes closed and explicit feedback conditions. The results showed that although explicit feedback is provided, the CoM sway induced by the sensory manipulations was not completely effaced. These results highlight the potential of using visual feedback to assess and train balance, since the use of feedback (up-weighting of visual inputs) on balance assessment cannot overcome and therefore mask sensory deficits.

In **Chapter III,** a mediolateral balance assessment tool was introduced (MELBA), which uses sinusoidal predictable and multisine unpredictable tracking tasks with the centre of pressure (CoP) as feedback on performance. This method quantifies performance using a transfer function between the target signal as input and CoP as output, from which phase-shift (response delay), gain (amplitude difference) and coherence (linearity) are calculated. Balance performance was then quantified using performance descriptors, defined as the frequency at which performance drops below a threshold of 90° for the phase-shift and 0.5 for the gain (f_{PS} and f_G) and the average phase-shift and gain defined by these bandwidths (PS_{mean} and G_{mean}).This first study using MELBA explored some methodological properties such as reliability and learning effects. The results of this study showed that MELBA's performance descriptors are reliable measures of mediolateral balance control. Small but consistent learning effects were found and suggest that similar methods can be used in training tools aimed to prevent balance impairment.

The CoM is the controlled variable in balance control and the ability to control and displace the CoM through CoP tracking assumes that consistent CoM movements would be elicited. In **Chapter IV,** this assumption was further explored by determining the amount of CoM displacement when using a CoP tracking tasks. Additionally, the benefits of using CoM tracking and changes in kinematic strategies between the two types of feedback (CoP and CoM) were explored. The results showed that CoP feedback in MELBA elicits consistent CoM displacements. However, different kinematics are employed when using CoM feedback, with more trunk movement and an ankle-to-hip shift as frequency increases. Hence CoM feedback may be preferable over CoP feedback despite the larger measurement effort involved.

The study presented in **Chapter V** was aimed to determine (1) the effect of age on and (2) the reliability of performance when using the new version of MELBA (CoM tracking). Balance performance on MELBA was compared between young (26±3 years) and older adults (72±5 years). A subsample of older and all young adults were reassessed to determine reliability of balance performance descriptors. In addition, all older adults performed a series of clinical balance tests and conventional posturography was done in another subsample. Results showed that bandwidths were lower in the older adults and the averaged values for phase-shift and gain within these bandwidths were also lower than in young adults. Performance descriptors showed good to excellent reliability in both groups and exhibited small but systematic between-session differences, indicative of learning. In contrast, all clinical scores were close to the maximum and no age effect was found using posturography. It was concluded that MELBA is a reliable tool to assess mediolateral balance in young and older adults, that the ability to accurately perform mediolateral CoM tracking deteriorates with age and that MELBA is more sensitive to these age-related impairments than posturography and clinical tests.

The ecological validity of MELBA was studied in **Chapter VI.** In this study, the MELBA balance performance descriptors were used to predict mediolateral stability during treadmill and daily-life walking in a group of community-dwelling older adults. To assess gait stability, the local divergence exponent was calculated using mediolateral accelerations obtained with an inertial sensor when walking on a treadmill for 5 minutes and in daily-life for one week. Results showed that the phase-shift bandwidth for the predictable target was significantly correlated with stability during treadmill walking, whereas the phase-shift bandwidth for the unpredictable target was significantly correlated with stability in daily-life gait. These correlations suggest that whereas performance in the unpredictable task can reflect the abilities to cope with unexpected disturbances occurring in daily-life, the predictable task may reflect the physical capabilities related to mediolateral balance control required to walk in unperturbed contexts. It was concluded that MELBA is an ecologically valid tool for mediolateral balance assessment in community-dwelling older adults who exhibit subtle balance impairments.

Overall, this thesis introduces a new balance assessment tool (MELBA), which was shown to be reliable, direction specific, challenging, to have no ceiling effects, and greater sensitivity to age than conventional balance measures. These features allow for identification of subtle balance impairments in older adults, which were predictive of instabilities in real-life gait. The findings of this thesis are further discussed in **Chapter VII** where limitations, clinical implications as well as directions for future research are presented.

Español

La introducción general en el **Capítulo I**, presenta una visión generalizada sobre la importancia de preservar la capacidad de mantener el balance en los adultos mayores, especialmente en la dirección mediolateral. El deterioro del balance ha sido señalado como una de las principales causas de las caídas relacionadas al envejecimiento, las cuales conllevan serias consecuencias para la movilidad independiente así como también para la calidad de vida. El deterioro de los sistemas sensorial y motor así como también el deterioro cognitivo afecta negativamente el proceso de integración sensorial el cual finalmente afecta la capacidad de un sujeto de enfrentar los desafíos físicos de la vida diaria. Sin embargo, las herramientas tanto clínicas como de laboratorio con las que hoy se cuenta para evaluar el balance presentan un "efecto techo" cuando se usan para determinar deterioros leves en adultos mayores que viven en la comunidad. Esto se debe posiblemente a que estas evaluaciones no presentan desafíos suficientes al sistema de control del balance como ocurre en situaciones de la vida diaria. Esto es de gran relevancia ya que herramientas sensitivas diseñadas para la detección temprana del deterioro del balance en los adultos mayores, pueden servir para identificar a aquellos que se podrían beneficiar con intervenciones para mantener o prevenir mayor deterioro. Además la evidencia señala que el balance mediolateral es de gran importancia debido a que la habilidad de transferir carga en las extremidades inferiores tiene un rol importante en las caídas en los adultos mayores cuando están de pie, suben escaleras o cuando caminan.

El **Capítulo II** presenta una investigación que tiene por objetivo determinar si al existir retroalimentación visual del centro de masa (CdM) los sujetos pueden sobrellevar la introducción de información sensorial errónea en el sistema de control del balance mediante la utilización de diferentes modalidades de manipulación sensorial. Este estudio utilizó manipulaciones de los sistemas vestibular (estimulación galvánica), propioceptivo (vibración musculotendinosa) y somatosensorial (estimulación nerviosa transcutánea) durante pruebas posturales estáticas con los ojos abiertos, cerrados y con retroalimentación visual. Los resultados muestran que a pesar de presentar retroalimentación visual explícita, el movimiento del CdM inducido por las manipulaciones sensoriales no puede ser completamente eliminado. Existe también una respuesta incrementada en las frecuencias contenidas en las señales de manipulación sensorial las cuales son más extensas cuando se usan señales de estimulación más complejas en los sistemas vestibular y propioceptivo. Estos resultados resaltan el potencial uso de retroalimentación visual para evaluar y entrenar el balance ya que su uso no oculta déficits sensoriales mediante el incremento en la ponderación de información visual.

El **Capítulo III** presenta una herramienta de evaluación del balance mediolateral (MELBA) la cual utiliza pruebas de seguimiento de objetivos que se mueven en forma sinusoidal y predecible y multisinusoidal e impredecible usando el centro de presión (CdP) como retroalimentación. Este método cuantifica el rendimiento mediante una función de transferencia entre el objetivo a seguir (señal de entrada) y el CdP (señal de salida) con la cual se calcula el cambio de fase (retardo de la repuesta), la ganancia (diferencia en la amplitud) y coherencia (linealidad). Para el cambio de fase y la ganancia se definieron umbrales basados en el rendimiento máximo durante la prueba para determinar descriptores del rendimiento. Este primer estudio usando MELBA exploró algunos aspectos metodológicos de la prueba como confiabilidad y efectos de aprendizaje. Los resultados muestran que los descriptores en MELBA son mediciones confiables del control del balance mediolateral, el cual adicionalmente puede dar una mejor idea de los mecanismos de integración visomotora que controlan el balance y por otro lado, menores pero consistentes efectos de aprendizaje encontrados podrían ser potenciados para ser usados en métodos de entrenamiento diseñados para prevenir el deterioro del balance.

El CdM es la variable controlada durante la locomoción humana y se espera que cuando se evalúa con MELBA la habilidad de desplazar y controlarlo se imponga mediante la retroalimentación del CdP. El **Capítulo IV** estos elementos se exploran con más profundidad, para determinar la cantidad de desplazamiento del CdM durante las pruebas de seguimiento en MELBA usando retroalimentación del CdP. Además se exploró los beneficios de usar retroalimentación explícita del CdM y se compararon cambios cinemáticos utilizados durante las pruebas de seguimiento usando ambos tipos de retroalimentación (CdP y CdM). Los resultados muestran que MELBA usando CdP provoca un consistente desplazamiento del CdM. Sin embargo, diferentes estrategias cinemáticas son empleadas cuando se utiliza CdM las que involucran mayor movimiento del tronco y un cambio en el origen del movimiento desde los tobillos hacia las caderas a medida que el contenido de frecuencias a seguir incrementa. De esta manera, retroalimentar usando el CdM por sobre CdP , a pesar de los mayores esfuerzos involucrados, podría ser más conveniente.

El estudio presentado en el **Capítulo V** tuvo por objetivo determinar (1) el efecto del envejecimiento y (2) la confiabilidad de las mediciones de rendimiento utilizando la nueva versión de MELBA (retroalimentación del CdM). En este estudio se cuantificó el balance de un grupo de adultos jóvenes (26±3 años) y un grupo de adultos mayores (72±5 años) usando 2 descriptores de cambio de fase y 2 de ganancia. Estos descriptores son definidos como la frecuencia a la cual el rendimiento cae bajo el umbral de 90° para el cambio de fase y 0.5 para la ganancia (f_{PS} and f_G) y el promedio de los valores de cambios de fase y ganancia en el ancho de banda definido por los anteriores (PS_{mean} and G_{mean}). Una muestra de los adultos mayores y la totalidad de los adultos jóvenes, fueron reevaluados para determinar la confiabilidad de los descriptores del balance. Además, todos los adultos mayores fueron clínicamente evaluados con pruebas funcionales estandarizadas. Adicionalmente pruebas posturográficas convencionales se aplicaron a un subgrupo de adultos mayores. El cambio de fase y la ganancia cayeron por debajo del umbral predeterminado a frecuencias más bajas en los adultos mayores que en los jóvenes y con valores dentro de este rango más bajos que en los adultos jóvenes. Todos los resultados de las evaluaciones clínicas estuvieron cercanos al máximo y los efectos del envejecimiento no fueron encontrados cuando se usó posturografía. Las mediciones de rendimiento del balance mediolateral mostraron menores pero sistemáticas diferencias entre sesiones lo que es indicativo de efectos de aprendizaje. En conclusión, la capacidad de ejecutar tareas de balance y seguimiento mediolateral (MELBA) se deteriora con el envejecimiento. La nueva versión de MELBA usando tareas de seguimiento con el CdM es confiable para evaluar el balance tanto de jóvenes como de adultos mayores y es más sensitiva a los efectos del envejecimiento que la posturografía y las evaluaciones clínicas.

En el **Capítulo V** se estudió la validez ecológica de MELBA. En esta investigación el rendimiento en MELBA (f_{PS}, f_G, PS_{mean} and G_{mean}) fue usado para predecir la estabilidad mediolateral durante la marcha, en cinta trotadora y durante la vida diaria, en un grupo de adultos mayores que viven en la comunidad (72±5 years). Para evaluar la estabilidad de la marcha, el exponente de divergencia local fue calculado usando aceleraciones obtenidas con sensores inerciales cuando los sujetos caminaron en una cinta trotadora (LDE_{TR}) y en la vida diaria (LDE_{DL}) durante una semana. El análisis utilizó correlaciones de Pearson para determinar correlaciones entre los descriptores de MELBA y LDE. Los resultados muestran que el ancho de banda del cambio de fase (f_{PS}) para el objetivo predecible (rango sobre -90º) fue significativamente correlacionado con LDE_{TR} mientras f_{PS} para el objetivo impredecible tuvo una correlación significativa con LDE_{DL}. Estas correlaciones indican que mientras el rendimiento en la tarea impredecible puede reflejar las habilidades para sortear perturbaciones inesperadas del balance que ocurren en la vida diaria, la tarea predecible refleja en mayor medida las capacidades físicas relacionadas al control del balance mediolateral para caminar en contextos más estables. En conclusión, MELBA es una herramienta ecológicamente válida para la evaluación del balance mediolateral en adultos mayores que viven en la comunidad quienes presentan menores grados de deterioro del balance.

En su conjunto, esta tesis presenta una nueva herramienta para la evaluación del balance (MELBA), la cual mostró ser confiable, de dirección específica, desafiante, sin efecto techo, de mayor sensibilidad al envejecimiento que mediciones del balance convencionales, capaz de cuantificar deterioros del balance menores en adultos mayores y de validez ecológica. Los hallazgos de esta tesis son discutidos en mayor profundidad en el **Capítulo VII** en donde las limitaciones, implicancia clínicas así como también la dirección que deben tomar futuras investigaciones son discutidas.

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A

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- **Cofré Lizama LE**., Pijnappels M., Reeves P., Verschueren S., van Dieën J. (2014) "The effect of aging on mediolateral balance control". *PLoS One (Accepted).*
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- Rispens SM, van Schooten KS, Pijnappels M., **Cofré Lizama LE.,** Daffertshofer A., Beek P., and van Dieën J. Gait characteristics associated with falls measured in the laboratory and in daily

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- **Cofré Lizama LE**., Pijnappels M., Faber G., Reeves P., Verschueren S., van Dieën J. "Can visual feedback efface the effects of sensory manipulations on mediolateral balance performance?" (Under Review).

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- **Cofré Lizama LE**, Faber G.H., Pijnappels M., Reeves P., Verschueren S., van Dieën J.H. The reliability and sensitivity of mediolateral balance assessment using a centre of mass tracking task (MELBA). 7th World Congress of Biomechanics (WCB 2014). July 6-11, 2010 Boston, USA.
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- **Cofré Lizama LE**, Pijnappels M., Reeves P., Verschueren S., van Dieën J., "Sensitivity of a mediolateral balance assessment tool to sensory manipulations". ISB 2013 Congress XXIV, Proceeding Abstract, Natal, Brazil.
- **Cofré Lizama LE**, Pijnappels M., Reeves P., Verschueren S., van Dieën J., "The effect of sensory manipulations on mediolateral balance performance". ISPGR 2013 Proceeding of the International Society for Posture and Gait Research, Akita, Japan.
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- **Cofré Lizama LE**, Lythgo ND. Ankle plantar-flexor power in human gait. ISB 2011 Congress XXIII, Proceeding Abstract, Brussels, Belgium.
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Academic Prizes/Scholarships

- **2014** Student travel Award, 7th World Congress on Biomechanics, Boston, USA.
- **2014** Best Oral Presentation, 3rd MOVE-AGE Annual Meeting (June 2014), Amsterdam, The Netherlands.
- **2013** Travel Grant, XXIV International Society of Biomechanics, Natal, Brazil.
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- **2010** Student travel Award, 6th World Congress on Biomechanics, Singapore.
- **2008** Bicentennial Scholarship to study a MSc in Australia (CONICYT, Chile).

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